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THE KINEMATIC CONSEQUENCES OF INVARIANT DYNAMICS IN CHILDREN 6 TO 18 YEARS OF AGE

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Abstract

The effect of limb dynamics on trajectory formation is unclear. The natural frequency of a limb is the major factor in its dynamics. It has previously been shown with an indirect measurement method that the natural frequency of body segments is invariant during human growth from the age of 6 to 18. The aim of our study was to determine, using a direct measurement method, whether human growth affects: (1) lower limb dynamics (i.e. the natural frequency of the lower leg) and (2) the maximum velocities of the knee during selected motor tasks. In 20 non-disabled children, 6–18 years of age, measurements were taken of the natural frequency of the lower leg (including the foot), and the maximum velocities of knee flexion and extension during voluntary movement (MVV) and at initial and terminal swing phases of self-paced walking (WAL). The velocities were also estimated using a dynamic model and the results were compared to the measured velocities with a paired t-test. Correlations among the frequencies, velocities, and body height (an indicator of growth) were calculated. The natural frequency of the lower leg (mean $\pm$ sd, $\omega_0 = 6.58 \pm 0.54 \text{ s}^{-1}$), maximum velocities of knee extension and flexion during voluntary movement, ($MVV_e = 10.1 \pm 1.8 \text{ rad s}^{-1}$ and $MVV_f = 7.8 \pm 1.3 \text{ rad s}^{-1}$, respectively), and maximum velocities of knee flexion and extension during the swing phase of walking ($WAL_f = 5.4 \pm 0.6 \text{ rad s}^{-1}$ and $WAL_e = 6.3 \pm 0.87 \text{ rad s}^{-1}$, respectively) were each found to be independent of body height. The MVV measured velocities were 22% larger and WAL_f measured velocities were 25% smaller than the velocities predicted from the dynamic model ($p < 0.05$). The study found that a segment's dynamic properties, as well as selected kinematics, may be considered invariant with human growth.

Keywords

dynamics; similarity; children; growth; velocity

1. Introduction

The planning and execution of human movement tasks require utilization of hundreds of structural elements of the motor system. The resulting movement in space and time (i.e. kinematics) of a joint is affected by its dynamic properties, its segment's interaction with the environment, and time-dependent control variables. The dynamic properties in a single

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biomechanical degree of freedom are described by the segment's mass and mass distribution, and the viscoelasticity of the joint, muscles and surrounding tissue. The relationship between the dynamics and kinematics about a joint is defined by an equation of motion, the basic form of which relates the torque (T_q) generated as a result of activation (a) of muscles acting about the joint to: segment motion (α), velocity (α'), acceleration (α''), segment mass (m), radius (r) from center of segment mass to the center of rotation, gravity (g), moment of inertia (J), joint stiffness (K) and damping (B).

$$J \alpha'' + B \alpha' + K \alpha + m g r \sin(\alpha) = T_q(a, \alpha, \alpha') \quad (\text{equation 1})$$

The relationships between kinematic parameters have been extensively investigated (e.g., Freund and Büdingen, 1978; Lestienne, 1979; Bizzi et al., 1982) during various movement tasks with respect to the complex control mechanisms responsible for them. The remarkable invariance commonly observed in kinematic patterns implies the utilization of complex control rules in trajectory formation. Applications of optimization criteria such as “minimum jerk” (Hogan, 1984) and “minimum effort” (Hasan, 1986) along with various kinematic or dynamic control variables (Stein, 1982) showed agreement with some experimental data. There is an increasing number of arguments supporting the view that movement trajectories may emerge as a result of dynamic system interactions with the environment through control networks (e.g., Bullock and Grossberg, 1988; Schöner, 1990; Feldman, 1995; Krylow and Rymer, 1997; Hof, 2001).

The human body contains approximately 280 biomechanical degrees of freedom, which makes the relationship between body dynamics and kinematics quite complex (Morecki et al., 1984). This relationship becomes further complicated when human development is considered. During the first 18 years of life, human body size increases by approximately 5 times, and the maturation of the nervous system occurs. There are two well-distinguished phases of body growth: (1) the first 5 years, when growth of most body parts and segments are not proportional to body height and (2) from 5 to 18 years, when growth of body segments is proportional. It was shown that highly variable control patterns in infants lead to the emergence of somewhat optimal (“adult-like”) patterns in older children. (Forssberg and Nashner, 1982; Hadders-Algra and Forssberg, 1998; Schneider et al., 1990; Holt et al., 2007). During the second phase, body growth can be considered almost geometrically similar (Lebiedowska et al., 2008). Control adaptation mechanisms have been observed during the first phase of growth, but it remains unclear whether similar mechanisms play a role during the second phase of growth (Bullock and Grossberg, 1988).

The dynamic behavior of a lower limb system is described primarily by the natural frequency (ω_0), which may be calculated directly from the period of free limb oscillations (T) using the

relationship: $\omega_0 = \frac{2\pi}{T}$. It may also be calculated indirectly as the square root of a ratio between the passive stiffness (K) and the moment of inertia (J) by the equation $\omega_0 = \sqrt{K J^{-1}}$, when the influence of gravitation may be neglected, or by the equation $\omega_0 = \sqrt{K J^{-1} + mgr \sin \alpha J^{-1}}$ when gravitation plays a role. Discrepancies between these methods of calculating natural frequency have been observed in previous studies, (Stein et al., 1996; Lebiedowska and Fisk, 1999).

The aim of our study was to determine, using a direct measurement method, whether human growth affects: (1) lower limb dynamics (i.e. the natural frequency of the lower leg) and (2) the maximum velocities of the knee during selected motor tasks. It was hypothesized that if limb dynamics are the major factor determining kinematics, then the maximum velocities achieved by children during selected voluntary motor tasks could be predicted from the

dynamics. In order to achieve this aim, we measured the natural frequency of the lower leg, the maximum voluntary knee flexion and extension velocities in a seated position and the maximum knee flexion and extension velocities during the swing phase of walking. Some of the results have been previously reported in abstract form (Lebiedowska, 2001).

2. Materials and methods

Twenty children (11 girls and 9 boys) of ages 5.7–18.6 years were evaluated. Their body height range was 133–175 cm and their body mass range was 23.6–74.6 kg. Prior to the experiments, the children were examined to confirm that they were non-disabled. All parents and children were informed about the procedures and aims of the study, and signed a written consent form approved by the Springfield Committee for Research Involving Human Subjects, Springfield, IL, USA.

Measures at the knee joint were obtained during one passive (i.e., with muscles non-activated) task and during two active (i.e. with muscles activated) tasks. For the passive task, the subject was seated and asked to relax while the lower leg was moved away from the neutral (resting) position and released. At resting position, the subject's lower limb was relaxed and approximately perpendicular to the floor. Knee angular motion was measured with an electrogoniometer (Biometrics, UK) (Figure 1) and the period of lower leg oscillation (T) was measured. Surface EMG signals were inspected from the rectus femoris and hamstrings muscle in order to confirm the relaxation.

For the first active task (MVV), the subject was seated and instructed to extend and then flex the knee as fast as possible several times at the available range of motion (ROM) from the resting position (subject relaxed, shank approximately perpendicular to the floor); full extension was considered to be 0 degrees. The range of motion (ROM) was measured in knee flexion (α_f) and extension (α_e). For the passive and first active tasks, the raw motion data were collected and digitized at 133 Hz with a Digidata 1200 interface (Axon Instruments, USA). Further calculations were performed using MATLAB 5.0 software (Math Works, Inc. USA). For the passive task, the natural frequency of the lower leg (including the foot) was calculated

using the period of oscillations according to the equation: $\omega_0 = \frac{2\pi}{T}$ (Figure 1). For the MVV task, velocity was calculated as the first derivative of angular knee motion achieved during knee extension and flexion (MVV_e and MVV_f respectively) using the diff.m routine of Matlab 5.0 software (Math Works, Inc. USA).

For the second active task (WAL), the subject was asked to walk at self-paced velocity with retro-reflective markers attached to standard body landmarks (Hallett et al. 1993). The maximum velocity of knee flexion (WAL_f) and knee extension (WAL_e) in the sagittal plane during the swing phase was measured with a 5-camera 3D TV Vicon system (Oxford Metrics, UK). The knee range of motion (α_{wal}) and maximum velocity in knee flexion (WAL_f) and extension (WAL_e) during the swing phase of walking were calculated using AMASS software (Adtech, USA).

For each of the three tasks, the experimental data were compared to data predicted using two simplified models. The first model predicted the natural frequency of the lower leg and a second one predicted the maximum knee velocities obtained during voluntary tasks. The predicted natural frequency, ω_m , was calculated using a pure mass model of the foot and lower leg segment having length l (referred throughout the paper as the “mass model”) with the equation:

$$\omega_m = \frac{3.32}{\sqrt{l}} \quad \text{equation (2)}$$

This equation was derived from the physical pendulum equation:

$$\omega_m = \sqrt{\frac{mgr_m l}{mr_g^2 l^2}} \quad \text{equation (3)}$$

where: m - segment mass

r_g - ratio of radius of gyration to length, l , of the segment ($r_g = 0.74$)

r_m - ratio of distance between centers of mass and rotation to length, l , of segment ($r_m = 0.61$)

l - 0.25 body height (Winter, 1990)

The maximum knee velocity was predicted based on the simplifying assumptions that the lower leg angular motion α followed a natural (sinusoidal) trajectory as it moved through the ROM (i.e., the system, described with ω_0 , had a sinusoidal trajectory response to a pulse input). Therefore,

$$\alpha = ROM \sin(\omega_0 t) ; \quad \text{equation (4)}$$

and the angular velocity α' was

$$\alpha' = ROM \omega_0 \cos(\omega_0 t). \quad \text{equation (5)}$$

Next, it was assumed that the maximum velocity occurred close to the initial position (i.e., for $t \rightarrow 0$, $\cos(\omega_0 t) \rightarrow 1$)

$$\alpha' = ROM \omega_0 \quad \text{equation (6)}$$

The maximum knee velocities were estimated as $MVV_e(\text{pred}) = \omega_0 \alpha_e$ and $MVV_f(\text{pred}) = \omega_0 \alpha_f$, for extension and flexion respectively. The maximum velocities during swing phase of walking were predicted at the same ROM (α_{wal}) as $WAL_{f,e}(\text{pred}) = \omega_0 \alpha_{\text{wal}}$. This model is referred to throughout the paper as the “dynamic model”.

Correlations and regression lines were calculated between ω_0 , MVV_e and MVV_f and body height, H (as the primary indicator of human body growth). A lack of statistically significant correlations would indicate that the particular measures could be considered invariant with respect to body height. Appropriate means and standard deviations were calculated. The measured and predicted velocities were then compared using a paired t-test. Statistical analysis was performed using Statistica software (Statsoft, Ltd.) and a statistical significance level of $p \leq 0.05$.

3. Results

The experimentally-derived natural frequency of the lower leg was invariant in children ages 6–18, ($\omega_0 = 7.3 - 0.6H$, $r = -0.19$, $p = 0.38$) with an average of $\omega_0 = 6.5 \pm 0.6 \text{ s}^{-1}$ [mean $\pm$ standard deviation]. The maximum knee velocities in extension ($MVV_e = 11.5 - 0.46 H$, $r = -0.04$, $p = 0.86$) and in flexion ($MVV_f = 4.9 + 2H$, $r = 0.24$, $p = 0.28$) were also independent of body height. The ranges of knee motion were $ROM_e = 1.34 \pm 0.22 \text{ rad}$ for knee extension and $ROM_f = 1.22 \pm 0.24 \text{ rad}$ for knee flexion. (Table 1).

During the swing phase of walking, the maximum knee velocity decreased slightly with increasing body height ($WAL_f = 6.64 - 0.83H$, $r = -0.27$, $p = 0.29$; and $WAL_e = 8.7 - 1.63H$, $r = 0.36$, $p = 0.15$), but the correlations were not statistically significant.

The natural frequency calculated from the mass model (Figure 3) decreased ($p < 0.001$) with increasing body height ($\omega_p = 7.47 - 1.44H$, $r = -0.66$). In the MVV tasks, the velocities predicted from the dynamic model were slightly lower than those measured, with a decrease of 22% during knee extension ($p < 0.05$), (Table 1). During walking, the predicted velocities from the dynamic model were lower than those measured, with a 25% decrease ($p < 0.05$) during knee flexion.

4. Discussion

4.1. Passive knee dynamics

In this study, the invariance of knee dynamics in children ages 6–18 has been supported using a direct measurement method. The natural frequency values of the leg obtained using the direct method were 50% smaller than those reported in a previous study that used an indirect method of measurement (Lebiedowska and Fisk, 1999). This may be due to a couple of factors: in the previous study, the gravitational component was not separated from the stiffness calculation; in addition, the small oscillation technique that was applied may have increased the stiffness values substantially due to stiffness' strong nonlinear behavior (e.g., short-range stiffness and/or thixotropy) (Hufschmidt and Schwaller, 1987; Lakie and Robson, 1988).

The natural frequency values observed in our study were similar to the values previously reported in 10-year-old children (Nordmark and Andersson, 2002). These authors reported a decrease of natural frequency with increasing age in young children between 2 and 10 years of age, and our study suggests that natural frequency may plateau between the ages of 6 to 18 years. Thus the major adaptation of motor patterns during growth in infants and young children (Forssberg and Nashner, 1982; Hadders-Algra and Forssberg, 1998), may not apply in older children.

The differences observed between the natural frequencies predicted from the mass model (ω_m) and those measured directly (ω_0) were due to the fact that the latter measure contains not only mass but also a stiffness component, ω_k :

$$\omega_0^2 = \omega_m^2 + \omega_k^2 \quad (\text{equation 7})$$

where $\omega_k = KJ^{-1}$

The stiffness component increases with body height to a large enough degree to offset the decline of natural frequency with age due to the pure mass component. This may be related to the increase in cross-sectional area (number of cross-bridges), length and pennation angle (Binzoni et al. 2001) during growth.

Our study was limited to one biomechanical degree of freedom; thus, the results may not be directly extended into complex movement patterns. It seems that, at least under some conditions, limb dynamics are invariant in children. Upon reduction of equation 1 (by division of J) and taking into consideration that the maximum isometric voluntary moment of force and moment of inertia are functions of fourth and fifth (respectively) power of body height, the following equation describes motion during knee flexion and extension:

$$\alpha'' + \beta\alpha + \omega_k^2\alpha + \omega_m^2\sin(\alpha) = \frac{tq(\alpha, \alpha')a}{H} \quad (\text{equation 8})$$

where $\beta = BJ^{-1}$ and $tq = TqJ^{-1}$

When our experimental data are incorporated (Figure 3), the equation becomes:

$$\alpha'' + \beta\alpha + (1.38 + 1.73H)^2\alpha + (7.8 - 1.69H)^2\sin(\alpha) = \frac{tq(\alpha, \alpha')a}{H} \quad (\text{equation 9})$$

For the central nervous system, the system under control does not change. Therefore, equation 9 indicates that, at least in some conditions (e.g., fast movements, movements in the horizontal plane), there is no need for motor control adaptation in knee activation patterns of growing children.

Analysis of equations 8 and 9 suggests that for unrestricted (through joint anatomy or environment) movements initiated at a particular level of activation, angular motion and velocity should decrease proportionally with increasing body height. As a consequence, the tangential velocity of the endpoint may not change.

4.2 Maximum velocity (MVV) task

During the MVV task, the subjects voluntarily moved their limbs as fast as possible from the resting position to full extension at the available ROM. The ROM for knee extension and knee flexion, respectively, was: $\alpha_e = 1.34 \pm 0.22$ rad and $\alpha_f = 1.22 \pm 0.24$ rad. During extension, the ROM was restricted by knee joint anatomy, and during flexion, the ROM was restricted by the examination table and the subject's seated position. The maximum velocity in knee flexion and knee extension did not decrease across subjects with respect to body height ($MVV_e = 10.7 \pm 2.17$ rad s⁻¹ and $MVV_f = 8.8 \pm 2.17$ rad s⁻¹, Table 1). Based on equations 8 and 9, this would suggest there was an increased level of activation (a) in the older (taller) children or improvement in performance at the muscle level (i.e. generation of a larger force at a given activation level). It is also possible that children executed the task at the same ROM, acquired through neural and biomechanical networks training at a younger age.

There are limited published data on maximum velocities in children. Compared to values reported in adults by Houston et al., 1988 (12.15 ± 1.5 rad s⁻¹), the values of the maximum knee extension velocity in our study were slightly lower; however, the ROM was smaller in our study, which could explain this discrepancy. Asai and Aoki (1996) compared the maximum tangential velocity of the hand in 6-year-old children to that in adults. They found that the velocity was approximately 2 times larger in the adults than in the children and suggested that this was due to performance improvement related to changes at the sarcomere level. However, the authors did not provide information about the angular velocity. If the angular velocity and ROM were the same in the adults as in the children, then the higher tangential velocity observed in the adults would suggest that they used the same level of muscle activation as the children. This is due to the fact that (a) the tangential velocity of a segment's end point is a product of

its angular velocity and the radius, and (b) the radius is directly proportional to the segment's length, which increases with body height.

For the MVV task, the maximum velocities predicted from a simple model based on dynamics (ω_0) and ROM corresponded well to those observed experimentally; in flexion (MVV_f), the difference was not statistically significant. This fact strongly suggests that dynamics play a major role in movement trajectory formation. Therefore, this simple model may be used to provide a rough estimation of kinematics given a system's dynamics.

4.3. Walking (WAL) task

The change of the knee joint angle during the swing phase of walking is almost entirely passive, except for at terminal swing and at initial stance. At terminal swing, the knee muscles co-contract to damp the knee oscillation that would occur at full extension. At initial stance, the leg muscles provide sufficient damping to dissipate the impact energy of the body and stiffness to prevent collapse of the body. Thus, angular velocity about the knee joint is a function of the joint's passive dynamics and the initial conditions/input. In our study, for a given ROM both WAL_f and WAL_e were independent of the subject's height. Since the knee's passive dynamics were invariant, it suggests that the same input was applied to propel the lower leg. In the simplified models used in this study, the initial position and momentum of the thigh were considered as major inputs resulting in angular motion of the knee (Grieve and Gear, 1966; Maillardet, 1977; Mena et al., 1981; Hoy and Zernickie, 1985; Machon and McMahon, 1980; Garcia et al., 1998). For the WAL task, the angular velocities predicted by the simple model corresponded well to those observed experimentally. In extension (WAL_e), there was no difference between the values; in WAL_f , the measured velocity was lower than that predicted by the dynamic model (Table 1), which may have been due to initial conditions that biased motion at the knee joint during swing. Further, the estimation used in our study was based only on the simple pendulum model. It is conceivable that a more realistic, double pendulum model incorporating the passive and active properties of uni- and bi-articular muscles acting around the knee and hip joints may explain the observed discrepancies.

Reports on freestanding seem to provide additional support to the validity of invariant dynamics during unrestricted postural movements. Forssberg and Nasher (1982) reported lower angular motion at the ankle in a 10-year-old child than in a 6-year old child. Berger et al. (1992) reported tangential acceleration of the head in 6-year-old children similar to that of adults. In addition the amplitude of the body center of pressure motion was reported as invariant in children 6–18 years old (Lebiedowska and Syczewska, 2000).

This study's results suggest that in non-disabled children from 6–18 years of age, a limb segment's dynamics are the major factors limiting the velocity of a joint. Agonist/antagonist muscle activation creates an input pulse to the limb system, and the limb follows its natural trajectory. This is consistent with the idea that invariant kinematics emerge from the interactions between the system and network rather than from pre-computed trajectories (Kelso et al., 1980; Bizzi et al., 1982; Bullock and Grossberg, 1988; Schöner, 1990; Feldman et al., 1995; Gottlieb et al., 1995; Krylow and Rymer, 1997; Hof, 2003; Holt et al., 2007) and that input pulse formation is the major factor propelling the segment (Gottlieb et al., 1995). The role that muscle properties play may be secondary, which may be why the extensive studies to find the correlation between the type of motor units (slow versus fast) and maximum velocity of movement (e.g., knee muscles vs. forearm muscles) have failed to find any relationship (Houston et al., 1988).

For unrestricted movement tasks, at a particular level of activation or at the same initial conditions, the angular motion and velocity of a body segment are inversely proportional to body height. Any diversion from this relationship may be attributed to factors such as

improvement in performance, higher levels of activation, and/or changes in muscle mechanics. Previous studies reported that growth of the human body between ages 6–18 years is almost geometrically similar (Lebiedowska et al., 2007); the current study presents arguments to extend geometrical similarity into some properties of body dynamics and kinematics.

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Figure 1.

Passive motion of the lower leg. The natural frequency (ω_0) of the lower leg was calculated from the period of oscillations (T) using the equation $\omega_0 = 2\pi T^{-1}$.

Figure 2. Knee motion and velocity during knee extension MVV_e ; (A) at maximum velocity, and (B) during the swing phase of self-paced walking.

Figure 3.

The natural frequency (ω_0) of the leg and its inertial and stiffness components (ω_m and ω_k) as a function of body height (H) in children 6–18 years old.

$$\omega_0^2 = \omega_m^2 + \omega_k^2; \text{ where}$$

$$\omega_0^2 = 46.04 - 3.07 H, r = -0.08$$

$$\omega_m^2 = 51.06 - 15.43 H, r = -0.67 \text{ and}$$

$$\omega_k^2 = 8.17 + 3.74 H, r = 0.10$$

g as a function of body height (H(m), horizontal axis) in children 6–18

Predicted		p	
Mean ± S.D.	Regression equation	Mean ± S.D.	
6.50 ±0.60	$g_m = 7.47 - 1.44H, r = 0.66^*$	5.17 ±0.41	0.0001**

Predicted			p
Mean ± S.D.	Regression equation	Mean ± S.D.	
8.80 ±1.73		10.7 ±2.17	0.009 **

Predicted			p
Mean ± S.D.	Mean ± S.D.		
7.80 ±1.80	8.02 ±1.57	<div><div><div>Regression equation</div><div>MTV₂(pred) = 8.40 - 0.43H, r = 0.05</div><div> </div></div></div>	0.68

$r = 0.24$

Predicted			p
Regression equation	Mean ± S.D.	Mean ± S.D.
$WAL_2 = 6.10 + 0.44H, r = 0.09$	6.26 ±0.80	6.70 ±0.80	0.10
$r = -0.36$	5.32 ±0.60	6.70 ±0.80	0.0001**

* statistically significant correlation
 ** statistically significant difference