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The effect of walking speed on the gait of typically developing children

Michael H. Schwartz[a,b,*], Adam Rozumalski[a,b], Joyce P. Trost[a]

[a]Gillette Children's Specialty Healthcare, St. Paul, MN, USA
[b]University of Minnesota, Minneapolis, MN, USA

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Abstract

Many gait studies include subjects walking well below or above typical self-selected comfortable (free) speed. For this reason, a descriptive study examining the effect of walking speed on gait was conducted. The purpose of the study was to create a single-source, readily accessible repository of comprehensive gait data for a large group of children walking at a wide variety of speeds. Three-dimensional lower extremity joint kinematics, joint kinetics, surface electromyographic (EMG), and spatio-temporal data were collected on 83 typically developing children (ages 4–17) walking at speeds ranging from very slow (>3 standard deviations below mean free speed) to very fast (>3 standard deviations above mean free speed). The resulting data show that speed has a significant influence on many measures of interest, such as kinematic parameters in the sagittal, coronal, and transverse planes. The same was true for kinetic data (ground reaction force, moment, and power), normalized EMG signals, and spatio-temporal parameters. Examples of parameters with linear and various nonlinear speed dependencies are provided. The data from this study, including an extensive electronic addendum, can be used as a reference for both basic biomechanical and clinical gait studies.

Keywords: Gait; Walking; Speed; Children; Kinematics; Kinetics; Electromyography; Spatio-temporal parameters

1. Introduction

It is well known that walking speed influences the fundamental elements of gait—joint rotations (kinematics), ground reaction forces, net internal joint moments and joint power (kinetics), muscle activity as measured by electromyography (EMG), and spatio-temporal parameters such as speed, stride length, and cadence (Murray et al., 1966, 1984; Paul, 1970; Andriacchi et al., 1977; Shiavi and Griffin, 1983; Kirtley et al., 1985; Frigo and Tesio, 1986; Shiavi et al., 1987, 1988; Detrembleur et al., 1997; Stansfield et al., 2001, 2006; Hof et al., 2002; Burnfield et al., 2004; den Otter et al., 2004; Segal et al., 2004; Diop et al., 2005; Hanlon and Anderson, 2006; Lelas et al., 2003; Grieve and Gear, 1966; van der Linden et al., 2002). The importance of walking speed is exemplified in Stansfield et al. (2001), where it was shown that speed, *not age*, was the primary determinant of kinematic and kinetic changes observed in growing children. The effect of speed on muscle activity is no less profound. Hof et al. (2002) examined EMG patterns at a variety of speeds, and found that the observed changes could be elegantly modeled by simple transformations on a small number of underlying patterns. At very slow speeds, den Otter et al. (2004) found systematic changes in EMG, but also noted the emergence of new muscle activation patterns, possibly due to increased stability demands.

The study presented here examines the effect of speed on kinematic, kinetic, EMG, and spatio-temporal data for a large number of typically developing children walking at a wide variety of speeds. Currently, multiple sources must be scoured to obtain this type of comprehensive data. Even when the data from each source is perfectly accurate, multi-source data synthesis presents difficult interpretive problems due to differences in subject characteristics, collection methods, biomechanical models, experimental conditions, walking speeds, and so on. Furthermore, few

*Corresponding author at: Gillette Children's Specialty Healthcare, 200 East University Avenue, St. Paul, MN 55101, USA. Tel.: +1 651 229 3929; fax: +1 651 229 3867.
E-mail address: schwa021@umn.edu (M.H. Schwartz).

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(if any) of these existing sources provide data in electronic format—making the task of using such meta-data even more daunting.

The existing literature is rich with studies analyzing the biomechanics of gait at different speeds. However, each of these studies leaves room for improvement in either design or data acquired and reported. For example, the study of Stansfield et al. (2001) does not include EMG data. In contrast, the studies of Hof et al. (2002) and den Otter et al. (2004) report EMG data, but lack joint kinematics and kinetics. Many of the previous studies have small numbers of subjects, or age and gender imbalances such as the pioneering work of Murray et al. (1984) examining seven adult women, or Kirtley et al. (1985) who studied 10 adult males.

A number of the past studies have employed a single or split-belt treadmill rather than level ground walking. The use of a treadmill allows precise speed control; however, it may also influence the gait mechanics (Bertram and Ruina, 2001). Since treadmills are impractical in many clinical applications (e.g. cerebral palsy, where balance and motor control are significantly compromised), over-ground data is often preferred as a reference. Thus, if data is to be used to aid in clinical decision-making—one of many possible uses—it is preferable to have data collected on level ground.

Due to design constraints, some of the previous studies, such as Stansfield et al. (2001), focused on a limited speed range. Others, like den Otter et al. (2004), focused only on slow speeds. However, it is typical to see a wide range of speeds within the clinical population. For example, in data from a comprehensive outcome study of children with Cerebral Palsy, subjects with a variety of gait pathologies were observed to walk at speeds ranging from 0.13 to 1.45 m/s (Schwartz et al., 2004).

It is also worth noting that many of the previous studies of speed effects focused on adults. Several studies did examine the effects of speed on the gait of children (Shiavi et al., 1988; Detrembleur et al., 1997; Stansfield et al., 2001, 2006; Diop et al., 2005; Grieve and Gear, 1966; Vaughan et al., 2003; Kimura et al., 2005; Hallemans et al., 2006; Chester et al., 2006). However, *most* of these focused on spatio-temporal parameters, as opposed to three-dimensional kinematics, kinetics, and EMG. There have been recent studies examining age-related differences in pediatric gait (Chester et al., 2006). There have also been studies examining differences between the gait of children and adults (Ganley and Powers, 2005, 2006). These studies found that while some parameters were affected by age—both within children, and between children and adults—most gait parameters did not differ. Taken together with the findings of Stansfield et al. (2001), and the current understanding of dynamic similarity and neuro-maturation, the effects of age seem to be of secondary concern (Vaughan et al., 2003; Hof, 1996; Kram et al., 1997; Donelan and Kram, 2000; Vaughan and O'Malley, 2005; Moretto et al., 2007).

This causes a practical dilemma. On the one hand, most gait data appears to be insensitive to age after gait has matured. Nevertheless, many clinical centers that use gait analysis to diagnose and treat movement disorders in children are more "comfortable" relying on pediatric data. For this reason, a pediatric reference is valuable to these centers; though the application of such data to adult studies is also legitimate.

The present study aims to enhance the existing body of literature describing speed-mediated effects on gait by presenting kinematic, kinetic, and EMG data for a large number of typically developing children walking at a wide variety of speeds. This study is not motivated by a lack of quality in the previous studies, but rather by the lack of a single comprehensive source for kinematic, kinetic, spatio-temporal, and EMG data at a wide range of speeds. The purpose of the study is *not* to explain the underlying biomechanics of the speed-mediated changes, but rather to provide a thorough report of the data in a single source. It will be left as a future task to explore the mechanisms responsible for the observed speed effects; including both the underlying biomechanical explanations and their clinical implications.

2. Methods

Ethical approval, parental consent, and subject assent were obtained prior to commencing data collection. Three-dimensional gait data was collected on 83 subjects who were given general instructions to walk at very slow, slow, self-selected comfortable (free), and fast walking speeds during a single testing session. The order of the speeds was not prescribed, other than collecting the free speed data first. Also, speed was not enforced (e.g. by a metronome or timer), as this may cause modifications to the gait pattern (Bertram and Ruina, 2001).

Motion data was collected using a 12-camera Vicon MX system (Vicon, Oxford, UK) operating at 120 Hz. The Vicon Plug-in-Gait marker set and model were used to generate the kinematic and kinetic data. In addition, a custom trunk model was used. This consisted of a marker over the 7th cervical spinous process (C_7), and symmetric markers approximately 2.5 cm to the left and the right of the sternal notch (L_{clav}, R_{clav}). The trunk segment coordinate system is analogous to that of the pelvis. Specifically, the first defining axis ($\approx$ mediolateral) is taken as the unit vector directed from R_{clav} towards L_{clav}. The second defining axis ($\approx$ dorsoventral) is perpendicular to the first axis, in a plane containing C_7, L_{clav}, and R_{clav}. The third axis ($\approx$ craniocaudal) is mutually perpendicular to the first two. For this study, trunk angles are reported relative to the pelvis using a Cardan angle sequence of sagittal→coronal→transverse.

Ground reaction forces were recorded using four force plates (AMTI, Watertown, MA), sampled at 1080 Hz. Consecutive force plate strikes of the left and right foot were acquired where possible, but individual force plate strikes were also allowed (this occurred primarily at slow and very slow speeds).

Surface EMG signals for the rectus femoris, medial and lateral hamstrings, anterior tibialis and medial gastrocnemius were acquired at 1080 Hz (Motion Lab Systems, Baton Rouge, LA). The SENIAM guidelines were followed for EMG acquisition (Hermens et al., 1999). Post-processing of the EMG data consisted of rectifying the signals and filtering with a zero-lag finite impulse response filter (3 Hz cutoff frequency). All trials for a subject-channel were then scanned for the peak signal (dynamic maximum). For each subject-channel, the signal was then normalized by this dynamic maximum in a manner similar to den Otter (den Otter et al., 2004). The resulting signals vary from 0 to 1.0, with 1.0 being the dynamic maximum. Generally, this dynamic maximum

Table 1
Subject characteristics

Group	Dimensionless speed range	Age (years)	Gender (F:M)	Height (m)	Mass (kg)
Very slow	<0.227	10.5 (3.6)	32:45	1.55 (0.19)	35.4 (10.4)
Slow	0.227–0.363	10.5 (3.5)	35:47	1.55 (0.19)	34.4 (9.6)
Free	0.363–0.500	10.5 (3.5)	35:48	1.56 (0.19)	34.1 (9.6)
Fast	0.500–0.636	10.4 (3.6)	32:44	1.55 (0.18)	35.4 (10.4)
Very fast	>0.636	10.0 (3.4)	26:25	1.49 (0.17)	38.3 (9.5)
Total	0.039–0.915	10.5 (3.5)	35:48	1.56 (0.21)	34.2 (9.6)

Age, height and mass displayed as mean (S.D.).

occurred at the subject's fastest speed, but this was not always the case. Normalization was performed to allow averaging of EMG data across subjects. It is recognized that this approach emphasizes activation *patterns* at the expense of signal *magnitude*.

Spatio-temporal parameters step length (SL), cadence (C) and walking speed (v) were rendered dimensionless using leg length (L_{leg}) following the scheme proposed by Hof (Hof, 1996):

$$\mathrm{SL}^* = \frac{\mathrm{SL}}{L_{\text{leg}}}, \tag{1}$$

$$C^* = \frac{C}{\sqrt{g/L_{\text{leg}}}}, \tag{2}$$

and

$$v^* = \frac{v}{\sqrt{gL_{\text{leg}}}}. \tag{3}$$

A randomly selected free speed trial was chosen from each subject, and the mean and standard deviation (S.D.) of the dimensionless speed from these free speed trials was computed. Five speed groups were defined:

$$\begin{array}{ll} \text{very slow} & 0 < v^* \leqslant \bar{v}^*_{\text{free}} - 3\,\mathrm{S.D.}_{\bar{v}^*_{\text{free}}} \\ \text{slow} & \bar{v}^*_{\text{free}} - 3\mathrm{S.D.}_{\bar{v}^*_{\text{free}}} < v^* \leqslant \bar{v}^*_{\text{free}} - 1\,\mathrm{S.D.}_{\bar{v}^*_{\text{free}}} \\ \text{free} & \bar{v}^*_{\text{free}} - 1\mathrm{S.D.}_{\bar{v}^*_{\text{free}}} < v^* \leqslant \bar{v}^*_{\text{free}} + 1\,\mathrm{S.D.}_{\bar{v}^*_{\text{free}}} \\ \text{fast} & \bar{v}^*_{\text{free}} + 1\mathrm{S.D.}_{\bar{v}^*_{\text{free}}} < v^* \leqslant \bar{v}^*_{\text{free}} + 3\,\mathrm{S.D.}_{\bar{v}^*_{\text{free}}} \\ \text{very fast} & \bar{v}^*_{\text{free}} + 3\mathrm{S.D.}_{\bar{v}^*_{\text{free}}} < v^* \leqslant \infty \end{array} \tag{4}$$

Each trial for each subject was then categorized, in a *post-hoc* manner, as belonging to one of these speed groups. Group averages were then computed for kinematics, kinetics, and EMG.

3. Results

3.1. Subject characteristics

Subjects in given speed groups were reasonably well balanced across genders, ages, and sizes (Table 1).

3.2. Spatio-temporal parameters

The *post-hoc* grouping resulted in a significant number of trials in each of the speed groups (Fig. 1). The dimensionless free speed group mean ($\bar{v}^*_{\text{free}} = 0.43$) and standard deviation ($\mathrm{S.D.}_{v_{\text{free}}} = 0.068$) were similar to previous studies, such as that of Vaughan, who reported a mean free dimensionless speed of $\bar{v}^*_{\text{free}} = 0.45$ (Vaughan et al., 2003). Cycle parameters steadily shifted towards shorter stance phase (68% very slow to 56% very fast) and shorter double support (36% very slow to 14% very fast) (Fig. 2). A step length versus cadence plot with speed isocurves revealed that the step length–cadence–speed pattern scaled uniformly (parallel isocurves), suggesting that these parameters maintain a fixed ratio above and below free speed (Fig. 3).

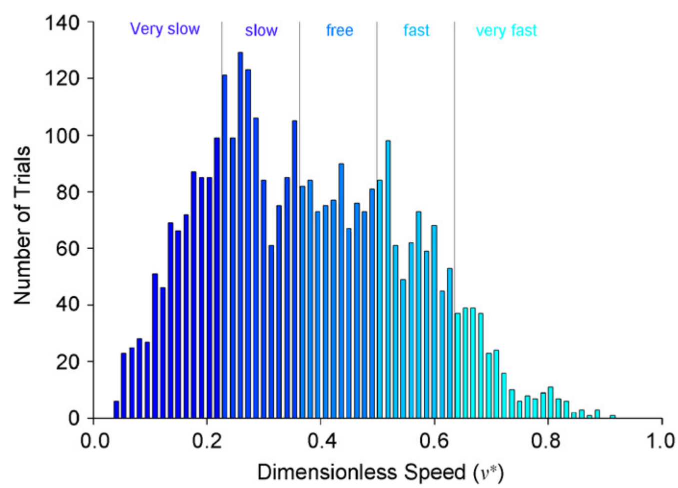

Fig. 1. Distribution of dimensionless walking speeds. *Post-hoc* classification resulted in trials that were well distributed among the five speed categories. Most subjects attained all speeds ($N_{\text{very slow}} = 77$, $N_{\text{slow}} = 82$, $N_{\text{SSC}} = 82$, $N_{\text{fast}} = 76$, $N_{\text{very fast}} = 51$), and every subject attained at least four speeds.

Due to the wide variety of data collected, and in keeping with the general descriptive, rather than explanatory purpose of the study, presentation of results will focus on general trends. Specific parameters will be presented to highlight the myriad of linear and nonlinear responses to speed that occur, and extensive data will be made available through the electronic addendum.

3.3. Kinematics

Sagittal, coronal, and transverse plane kinematics exhibited significant speed-related effects (Figs. 4 and 5(a)–(d)). The majority of local extrema scaled with speed in a variety

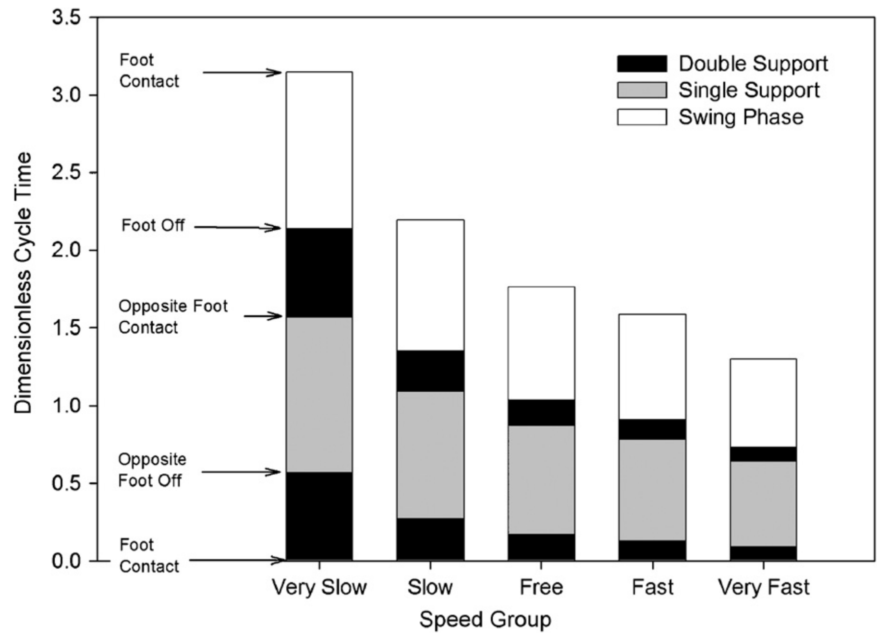

Fig. 2. Cycle events. Foot contact and foot off times for the ipsi- and contralateral leg are shown. Stance phase and double support both decrease steadily with increasing speed.

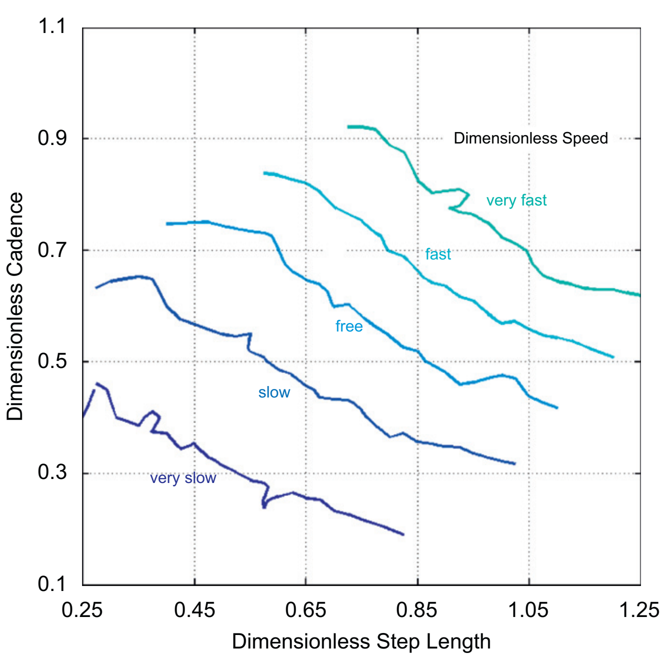

Fig. 3. Dimensionless step length, cadence and speed. The nearly parallel speed isocurves imply a constant relationship between these variables across a wide range of speeds.

of linear and nonlinear ways (Fig. 5(a)–(d)). Some parameters, such as hip flexion at foot contact, scaled nearly linearly (Fig. 5(a)).[1] In contrast, ankle dorsiflexion at foot off exhibited a plateau at low speeds (b), while maximum hip extension exhibited a plateau at high speeds (c), and pelvic tilt range-of-motion had a concave upward shape, increasing to either side of the slow speed group (d).

3.4. Kinetics

As was true for kinematics, kinetic data (ground reaction force, net internal joint moment, net joint power) also exhibited a variety of responses to speed (Figs. 6–8 and 9(a)–(d)). For example, while peak-braking force was nearly linear (a), peak propulsive force exhibited a plateau at high speed (b), maximum knee flexion moment in late swing exhibited a plateau at low speed (c), and peak hip power generation in early stance exceeded linear growth (d). The pervasiveness of the speed effect is extensive. Consider knee power (Fig. 8), which exhibits eight extrema, and is generally regarded by clinical gait analysts as no more meaningful than a spaghetti noodle carelessly dropped onto a set of axes. Despite this seemingly random appearance, six of the eight knee power extrema vary systematically with speed.

3.5. EMG

EMG signals were also found to be strongly affected by speed (Figs. 10 and 11(a)–(d)). The medial hamstrings activity at foot contact scaled linearly with speed (a). Rectus femoris in early swing showed a plateau at low speed (essentially inactive at $<10\%$ of maximum), but was quite active at high speed (b). In contrast, the early stance-phase activity of the anterior tibialis

[1] It should be noted that the difference between adjacent speed group means is nearly constant ($\delta_{\bar{v}^*} = 0.12 - 0.14$), allowing the categorical x-axis to approximate a scale axis.

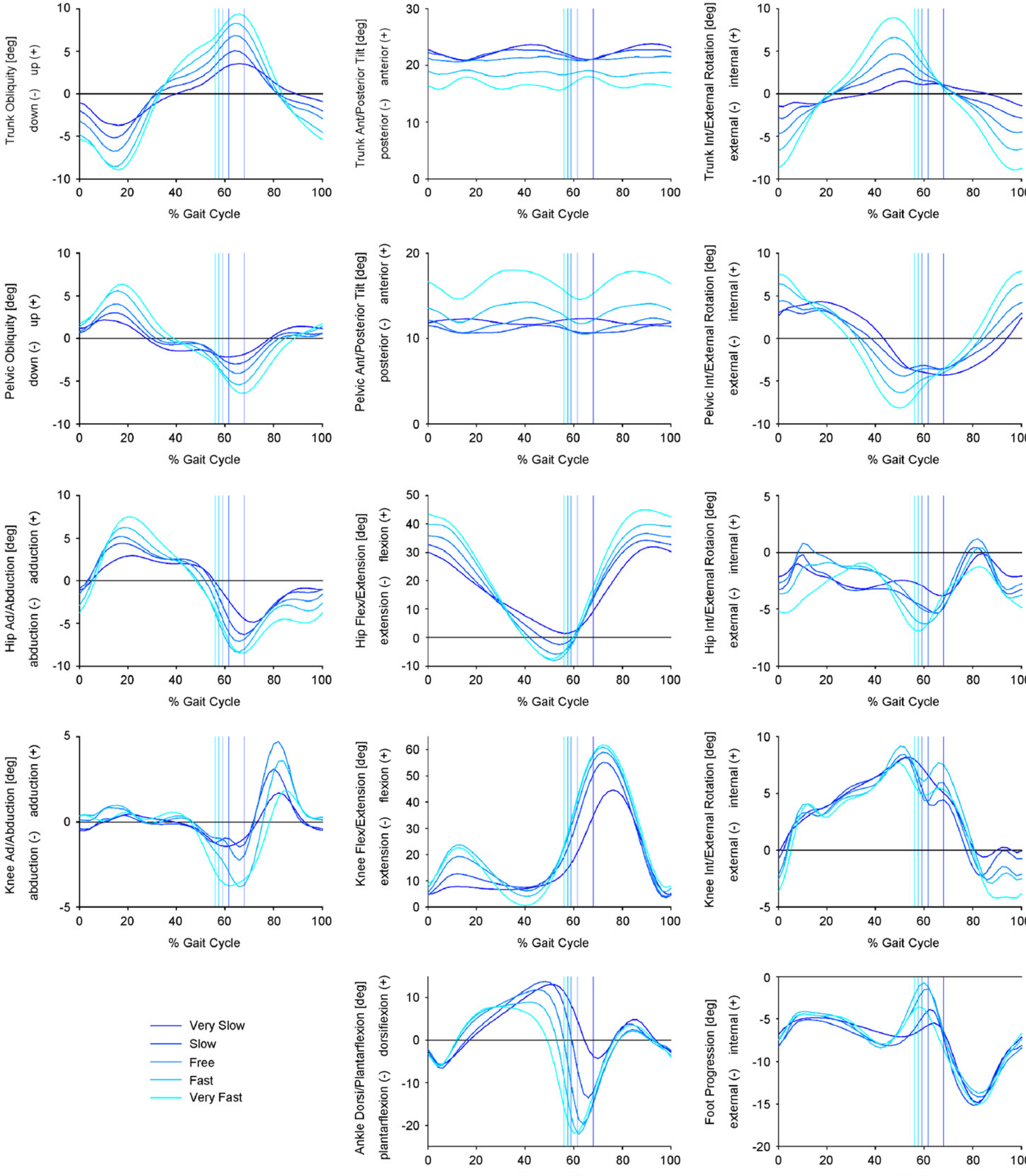

Fig. 4. Joint rotations. Three-dimensional joint rotations are shown. Each line represents the average of trials within the corresponding group. The color code (dark blue = very slow, through cyan = very fast) is maintained throughout the remaining kinematic, kinetic and EMG plots.

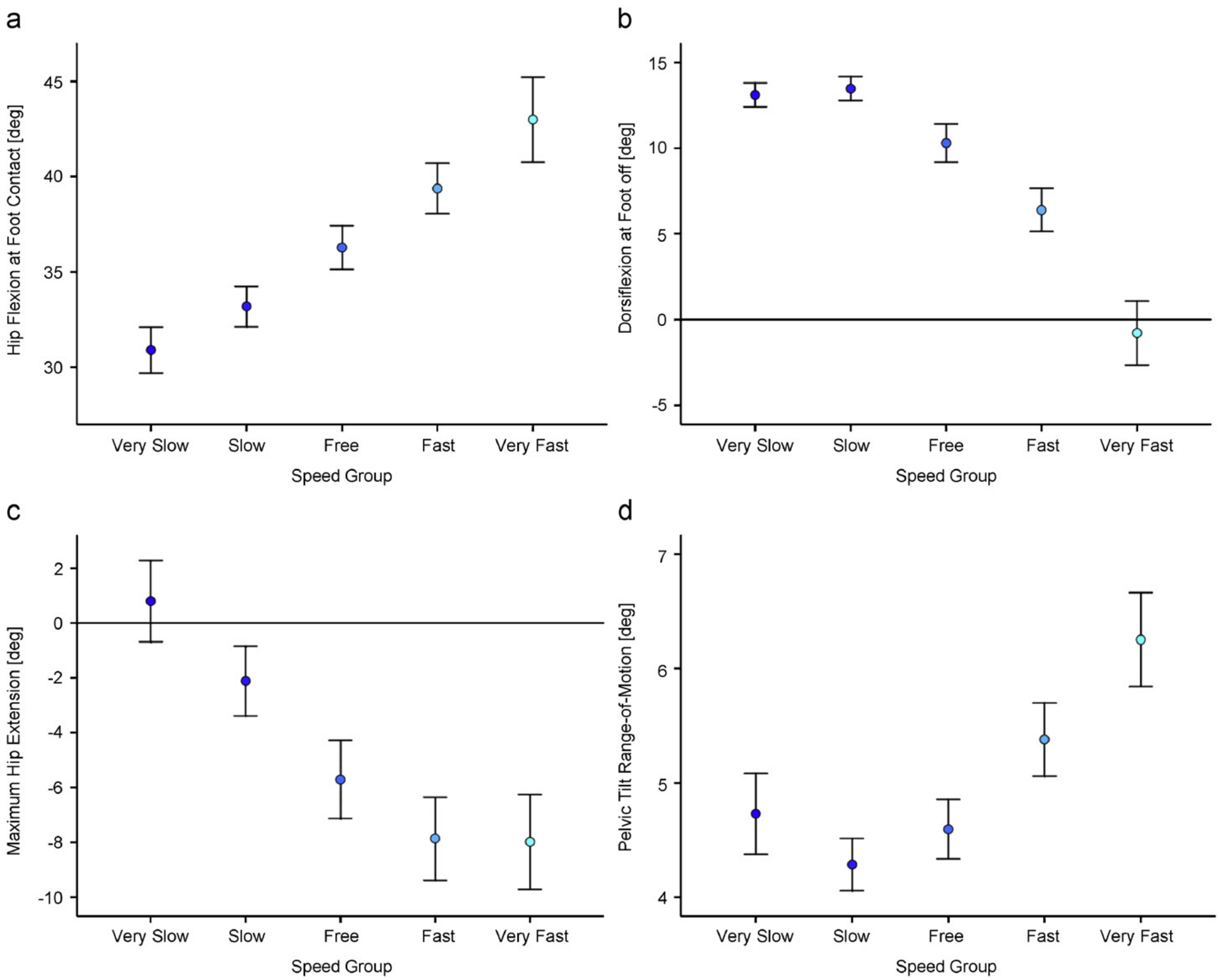

Fig. 5. (a)–(d) Examples of speed-mediated kinematic responses. Hip flexion at foot contact (a) various linearly with speed. Ankle dorsiflexion at foot off (b) exhibits a plateau at lows speeds, while maximum hip extension has a plateau at high speeds (c). The final plot, pelvic tilt range-of-motion (d) is concave upward, with a minimum at slow speed.

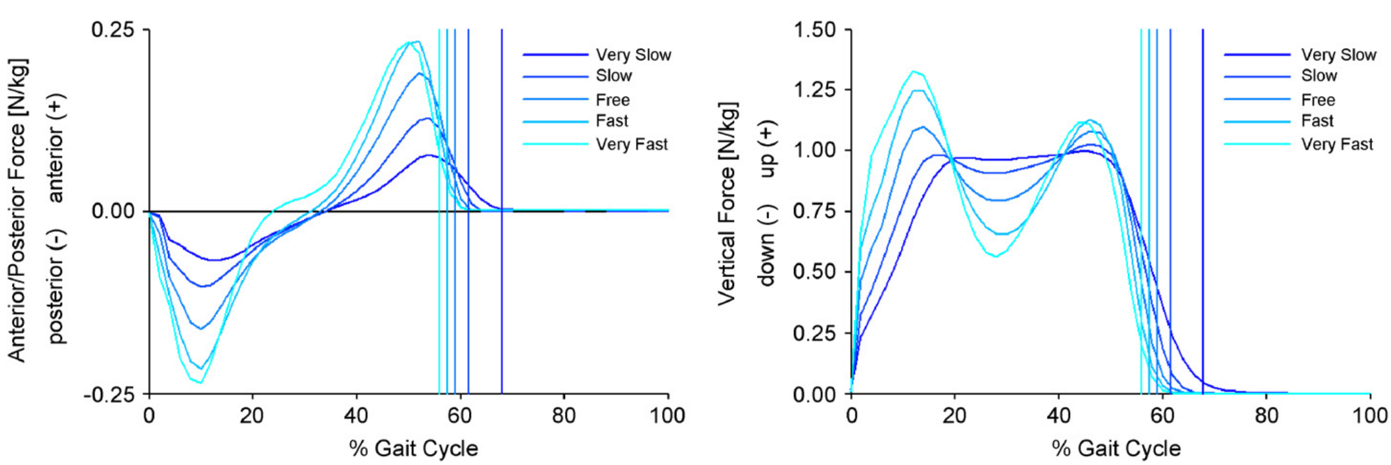

Fig. 6. Ground reaction forces. Anterior/posterior and vertical components of the ground reaction force are shown. Each line represents the average of trials within the corresponding group.

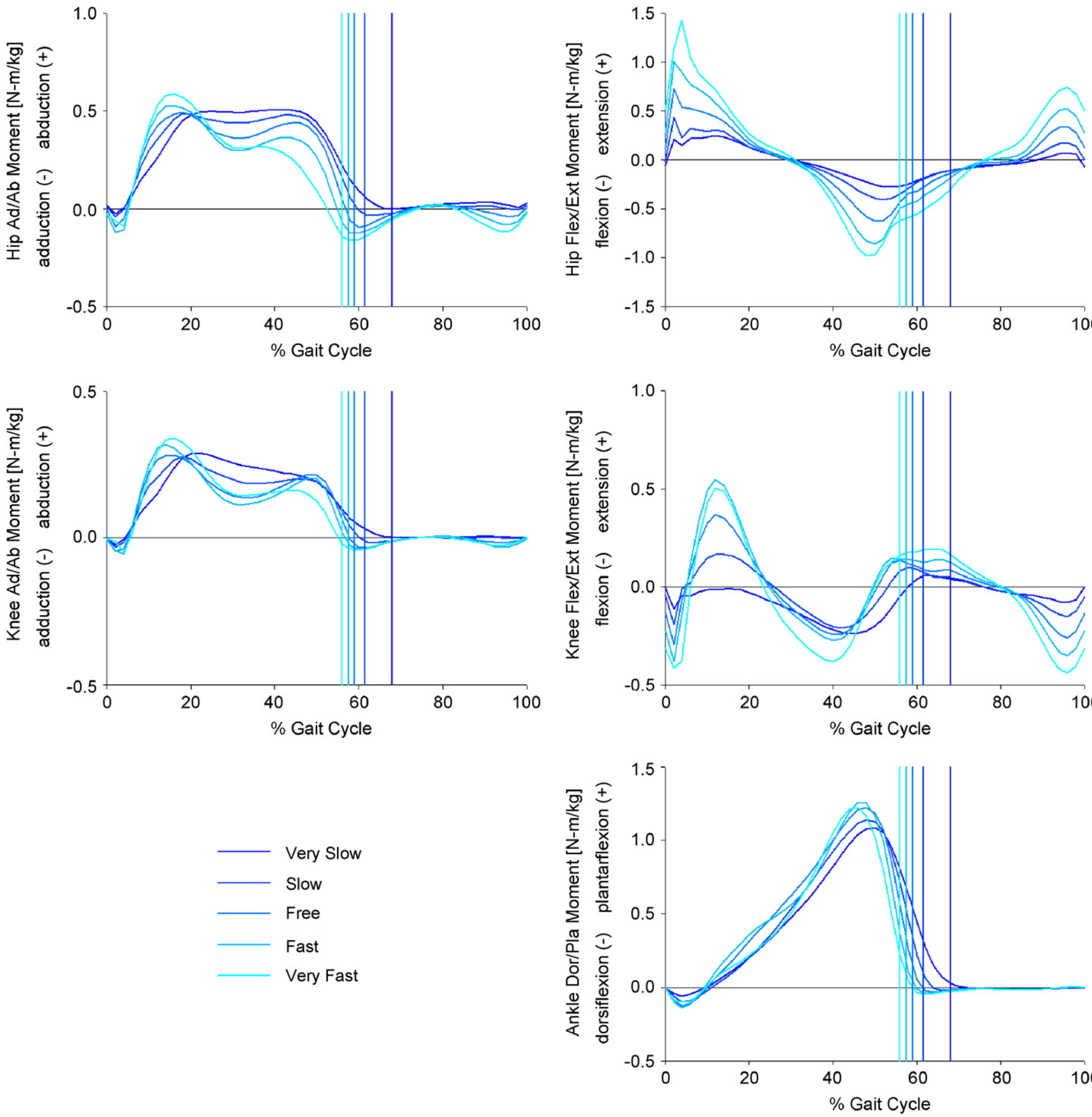

Fig. 7. Joint moments. Coronal and sagittal plane internal joint moments are shown. Each line represents the average of trials within the corresponding group.

responded linearly at slower speeds, but appeared to plateau at high speeds (c). Meanwhile, the early swing phase peak of the anterior tibialis exhibited a supra-linear growth (d).

Values of these and many additional extracted parameters (kinematics, kinetics, EMG, and spatio-temporal parameters) are available in the electronic addendum, along with an ANOVA table detailing significant differences in the parameters between speed groups.

4. Discussion

This study compiled a comprehensive set of gait data (kinematics, kinetics, spatio-temporal parameters, and

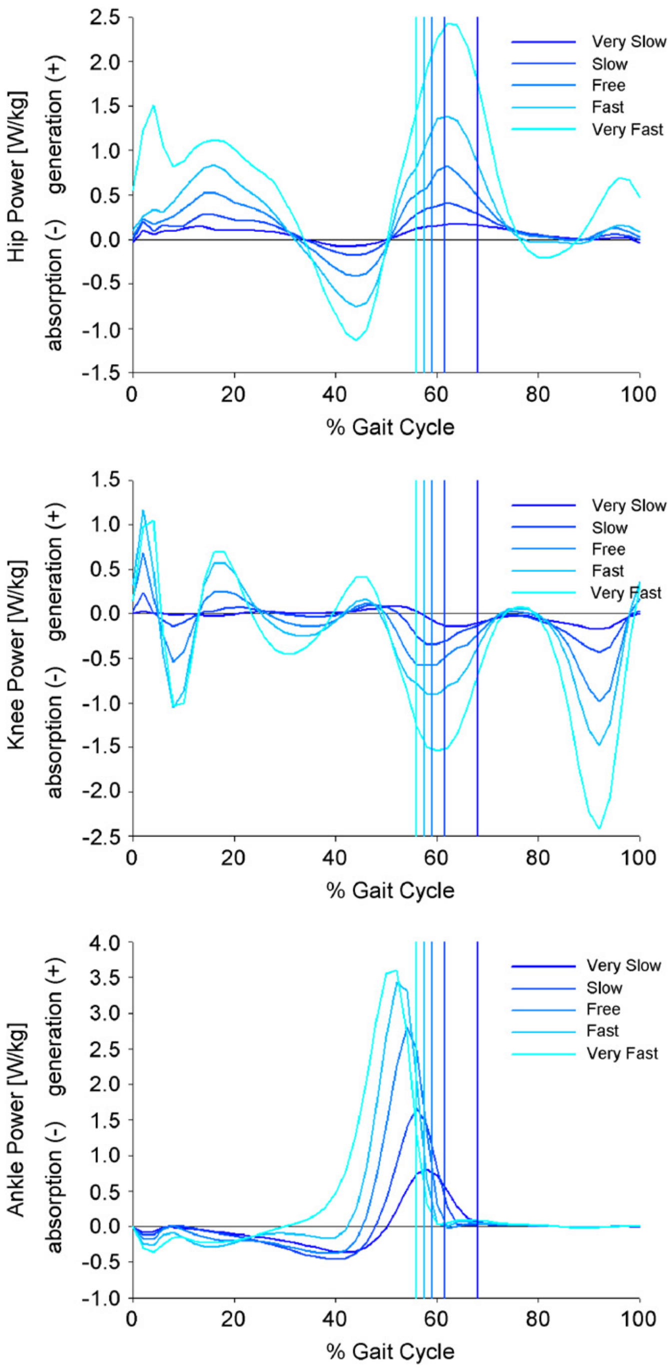

Fig. 8. Joint power. The total power generated/absorbed at the hip, knee, and ankle are shown. Each line represents the average of trials within the corresponding group.

surface EMG) on a comparatively large sample ($N = 83$) of children (age range 4–17) walking at a variety of speeds (dimensionless speed range 0.039–0.915, dimensional speed range 0.11–2.56 m/s). This data is meant to provide a thorough description of speed-related changes in gait of typically developing children. The group data (means and standard deviations) are available for download as an electronic addendum to this article. Possible uses of this data include both clinical studies (e.g. *how can one properly account for covariate effect of speed when assessing neuromuscular pathology*?) and basic biomechanical analyses (e.g. *what accounts for the change in the power ratio between the hip and knee at faster speeds*?).

While most of the data presented in this study could be gleaned from meta-analysis of the existing literature, there is no single source that supplies kinematics, kinetics, EMG, and spatio-temporal data together. As noted earlier, there are numerous analytic and statistical challenges posed by multi-source studies. Despite the differences in underlying models, methods, and samples, general comparisons to the previously published literature are enlightening. Changes in EMG, joint rotations, moments, and power were all primarily characterized by amplification of peak values with increasing speed. This is similar to previously published studies speed-related effects. Careful inspection revealed that some subtler findings from past work were also confirmed. For example, the reversal in phase of pelvic tilt at very slow speeds (Fig. 4) can also be seen in Fig. 2 of van der Linen (van der Linden et al., 2002). The amplification of EMG peaks from slow through fast was similar to those reported by Hof (Hof et al., 2002). For example, in Hof's study, the amplification of the medial gastrocnemius peak was about 1.6-fold (140–225 μV as manually digitized from Hof's Fig. 1(a)) for adults walking at a dimensionless speed range of 0.24–0.57. These speeds correspond to the low end of the slow and middle of the fast groups in the present study, where the amplification of medial gastrocnemius was found to be approximately 1.5-fold (45–67% maximum). It is noteworthy that Hof's data was collected on adults, while the data from this study was from children. Despite that fact, by matching dimensionless speeds, the data was found to be comparable.

It is worth noting that almost every aspect of gait appears to be sensitive to walking speed (spatio-temporal parameters, kinematics, kinetics, and EMG). Various parameters (e.g. selected local extrema) respond to speed in different ways—including linear and a variety of nonlinear patterns. Interpreting the individual responses—for example, understanding why peak braking force is linear, but peak propulsive force has a plateau at high speed—is beyond the scope of this study. Future studies by the present authors, or other authors choosing to access this data, may reveal the mechanics behind some of these observations.

There are certainly limitations that arise as a result of methodological choices. One of the main difficulties in studies of this type is how to vary speed. Here the choice was made to use vague instructions (''walk fast'', ''walk very slowly'') and *post-hoc* grouping. This resulted in a good distribution of speeds and a consistent step length–cadence–speed relationship (Figs. 1–3). With this approach, there is a risk that some subjects may not achieve each speed an equal number of times (i.e. that the data will be unbalanced). To some degree this turned out to be the case. However, most subjects attained all speeds

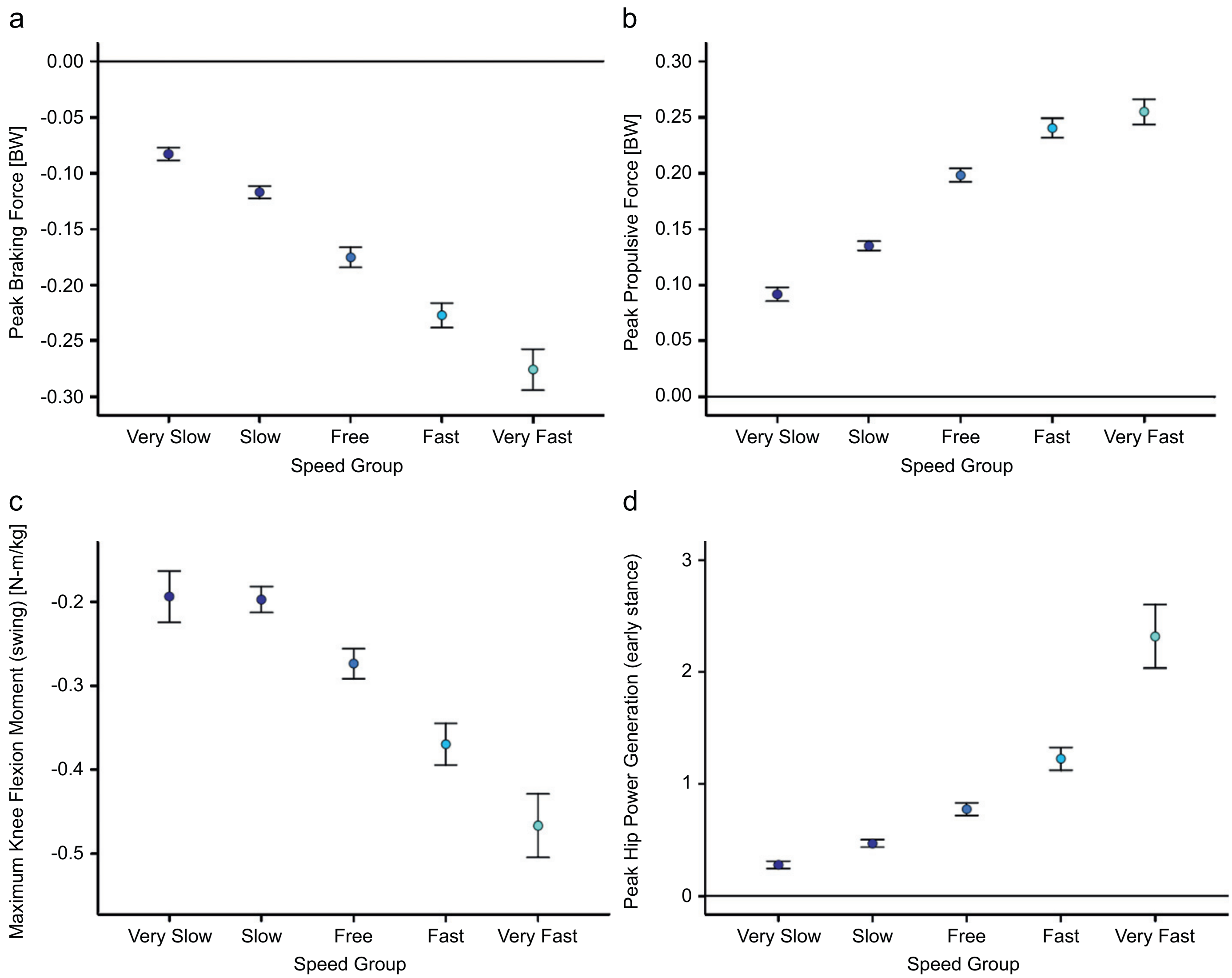

Fig. 9. (a)–(d) Examples of speed-mediated kinetic responses. Peak braking (rearward) ground reaction force varies linearly (a), while its forward counterpart (peak propulsive force) plateaus at high speeds (b). Maximum knee flexion moment has a plateau at low speeds (c). Peak hip power generation in early stance (sometimes called the H1 burst) increases at a greater-than-linear rate as speed increases (d).

($N_{\text{very slow}} = 77$, $N_{\text{slow}} = 82$, $N_{\text{free}} = 82$, $N_{\text{fast}} = 76$, $N_{\text{very fast}} = 51$), and each subject attained at least four speeds. It should be noted that subjects were never explicitly asked to walk "very fast", accounting for the drop-off in subjects producing trials in this category.

Another limitation of this study is the fact that this data was processed using the Vicon Plug-in-Gait model. This model is specific to one manufacturer, and contains numerous simplifications which may not be appropriate for some studies. For example, the hip joint centers are based on manufacturer specific anthropometric regression equations, rather than functional models. This results in both random and systematic errors in the hip center location (Leardini et al., 1999). Similar modeling compromises exist at the knee joint—where the flexion/extension axis is defined by an alignment device and the knee lacks translational degrees of freedom. Nevertheless, both the model and the underlying simplifications are common in the clinical gait community, and are representative of most conventional/commercial gait analysis models. Thus this data should have widespread applicability. As an aside, data was collected simultaneously with a functional model for hip and knee joint parameters (Schwartz and Rozumalski, 2005). The data derived from the functional model will be compared to the Vicon model in upcoming studies.

There is an extensive literature describing the effects of speed on gait. Thus, to some extent, the present study duplicates the efforts and data from prior studies. However, as noted earlier, each of the prior studies was lacking in at least one design or data element. It is possible to piece together information from multiple sources (i.e. EMG from one study, kinematics and kinetics from

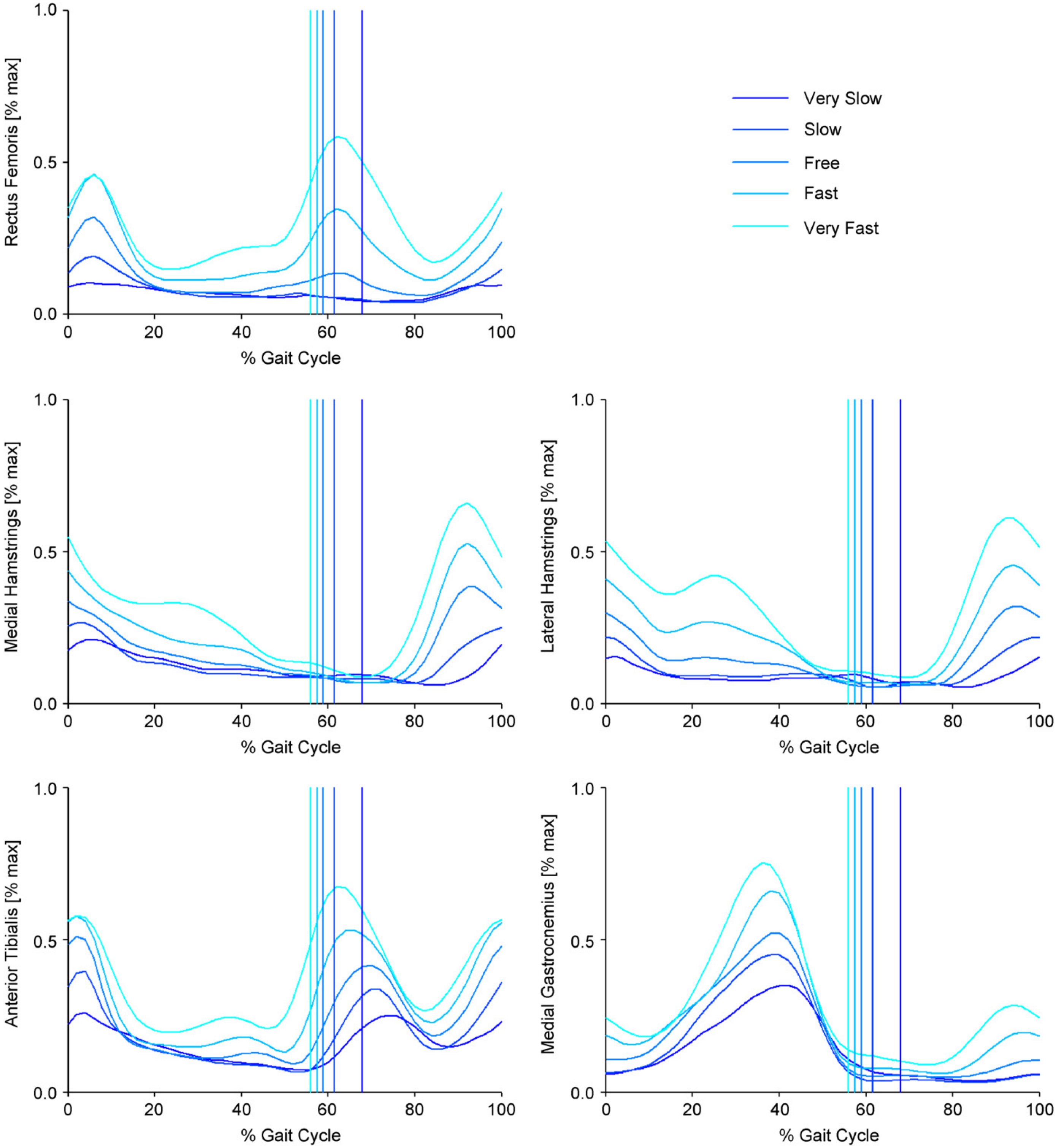

Fig. 10. Muscle activity. Surface EMG signals of several major muscle groups are shown. Each line represents the average of trials within the corresponding group.

another). However, due to differences in study designs, subject characteristics, biomechanical models, sampling methods, and equipment, conclusions from such meta-analyses are tenuous. Thus, it is worthwhile to have a single compiled source of data, as is provided in this study.

Conflict of interest

The authors listed on the manuscript had no conflict of interest when performing the study or when preparing the manuscript.

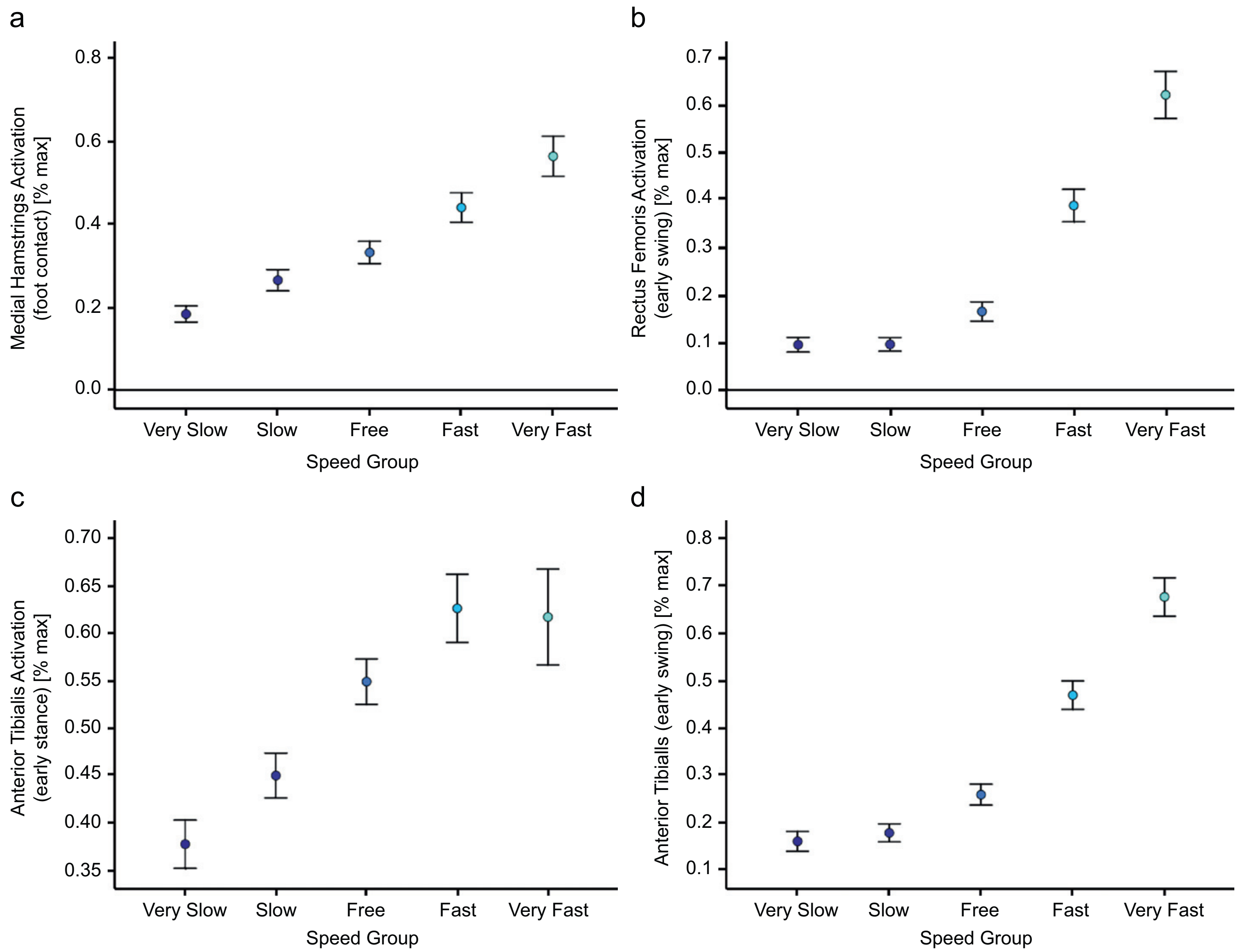

Fig. 11. Examples of speed-mediated EMG responses. The medial hamstrings activation at foot contact increases linearly with speed (a). The rectus femoris is essentially off at slower speeds, but increases rapidly at free speed and above (b). In contrast, the anterior tibialis activation in early stance exhibits a plateau at higher speeds (c), while the early swing phase activation of the muscle increases at a faster-than-linear rate with speed (d).

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Appendix A. Supplementary materials

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.jbiomech.2008.03.015.

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