Supplementary material

Supplementary material A: Spline fitting procedure

This appendix summarizes the reconstruction procedure of the reference trajectories. Between each consecutive pair of key-events a quintic spline is formulated. Each spline is based on 6 constrains (initial and final position, velocity and acceleration) and requires a fifth order polynomial:

$$\begin{aligned}
&S(x) = A_i + B_i x + C_i x^2 + D_i x^3 + E_i x^4 + F_i x^5 \\
&\frac{dS}{dx}(x) = B_i + C_i 2x + D_i 3x^2 + E_i 4x^3 + F_i 5x^4 \qquad \text{for } x_i \le x \le x_{i+1} \\
&\frac{d^2S}{dx^2}(x) = C_i 2 + D_i 6x + E_i 12x^2 + F_i 20x^5
\end{aligned} \qquad [1]$$

Where $S(x)$ represent the spline (between key-event i and key-event i+1) and A-F its coefficients. Filling in these equations for 2 subsequent key-events yields:

$$\begin{bmatrix} 1 & x_i & x_i^2 & x_i^3 & x_i^4 & x_i^5 \\ 0 & 1 & 2x_i & 3x_i^2 & 4x_i^3 & 5x_i^4 \\ 0 & 0 & 2 & 6x_i & 12x_i^2 & 20x_i^3 \\ 1 & x_{i+1} & x_{i+1}^2 & x_{i+1}^3 & x_{i+1}^4 & x_{i+41}^5 \\ 0 & 1 & 2x_{i+1} & 3x_{i+1}^2 & 4x_{i+1}^3 & 5x_{i+1}^4 \\ 0 & 0 & 2 & 6x_{i+1} & 12x_{i+1}^2 & 20x_{i+1}^3 \end{bmatrix} \bullet \begin{bmatrix} A_i \\ B_i \\ C_i \\ D_i \\ E_i \\ F_i \end{bmatrix} = \begin{bmatrix} x_i \\ \frac{dy}{dx}_i \\ \frac{d^2y}{dx^2}_i \\ x_{i+1} \\ \frac{dy}{dx}_{i+1} \\ \frac{d^2y}{dx^2}_{i+1} \end{bmatrix} \qquad [2]$$

Which can be written as:

$$\begin{bmatrix} A_i \\ B_i \\ C_i \\ D_i \\ E_i \\ F_i \end{bmatrix} = inv \begin{bmatrix} 1 & x_i & x_i^2 & x_i^3 & x_i^4 & x_i^5 \\ 0 & 1 & 2x_i & 3x_i^2 & 4x_i^3 & 5x_i^4 \\ 0 & 0 & 2 & 6x_i & 12x_i^2 & 20x_i^3 \\ 1 & x_{i+1} & x_{i+1}^2 & x_{i+1}^3 & x_{i+1}^4 & x_{i+41}^5 \\ 0 & 1 & 2x_{i+1} & 3x_{i+1}^2 & 4x_{i+1}^3 & 5x_{i+1}^4 \\ 0 & 0 & 2 & 6x_{i+1} & 12x_{i+1}^2 & 20x_{i+1}^3 \end{bmatrix} \bullet \begin{bmatrix} x_i \\ \frac{dy}{dx}_i \\ \frac{d^2y}{dx^2}_i \\ x_{i+1} \\ \frac{dy}{dx}_{i+1} \\ \frac{d^2y}{dx^2}_{i+1} \end{bmatrix} \qquad [3]$$

Where x_i represents the x parameter of the i[th] key-event. The values for the x , y , $\frac{dy}{dx}$, and $\frac{d^2y}{dx^2}$ parameters are calculated for a specific speed and body-height with the regression equations provided in table 1-4. For example: the x parameter of the 6 key-events for the knee joint trajectory are calculated according to:

$$\begin{bmatrix} x_1 \\ x_2 \\ x_3 \\ x_4 \\ x_5 \\ x_6 \end{bmatrix} = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 17{,}103 & 0 & 0 & 0 \\ 48{,}542 & -0{,}998 & 0 & 0 \\ 68{,}947 & -6{,}096 & 0{,}611 & 5{,}9067 \\ 85{,}816 & -4{,}480 & 0{,}519 & 0 \\ 92{,}489 & 0 & 0 & 0 \end{bmatrix} \bullet \begin{bmatrix} l & v & v^2 \end{bmatrix} \quad \underset{\substack{v = 3kph \\ l = 1.75m}}{\Rightarrow} \quad \begin{bmatrix} x_1 \\ x_2 \\ x_3 \\ x_4 \\ x_5 \\ x_6 \end{bmatrix} = \begin{bmatrix} 1.00 \\ 17.10 \\ 45.55 \\ 66.60 \\ 77.04 \\ 92.49 \end{bmatrix} \qquad [4]$$

[4]

Where v represents the walking speed and l the body-height. Here the x parameters are calculated for a walking speed of 3 kph and for a subject with a body-height of 1.75 m. The y, $\frac{dy}{dx}$, and $\frac{d^2y}{dx^2}$ parameters are calculated in a similar way (Table 3). To ensure continuity of the spline we define the angle, velocity and acceleration at the end of the sixth last spline to be equal to the start of the first spline:

$$\begin{aligned} x_7 &= 101 \\ y_7 &= y_1 \\ \frac{dy}{dy_7} &= \frac{dy}{dy_1} \\ \frac{d^2y}{dy^2{}_7} &= \frac{d^2y}{dy^2{}_1} \end{aligned} \qquad [5]$$

Filling in the obtained x, y, $\frac{dy}{dx}$, and $\frac{d^2y}{dx^2}$ parameters in a pairwise fashion in equation 3 yields the coefficients for the 6 splines.

$$1 \le x \le 17.1 \qquad S(x) = 6.67 \cdot 10^{-1} - 6.85 \cdot 10^{-1} x + 3.04 \cdot 10^{-1} x^2 - 1.45 \cdot 10^{-2}\ x^3 - 2.86 \cdot 10^{-5}\ x^4 + 8.45 \cdot 10^{-6} x^5$$

$$17.1 \le x \le 45.5 \quad S(x) = -\ 3.23 \cdot 10^1 + 7.72x - 4.57 \cdot 10^{-1} x^2 + 1.24 \cdot 10^{-2} x^3 - 1.70 \cdot 10^{-4}\ x^4 + 9.73 \cdot 10^{-7} x^5$$

etc.

Combining these splines creates the reference trajectory as shown in Fig. 4. The same approach is used to generate the trajectories for the others joint (table 1-4).

The method described above is also provided in matlab code (see “createRefTrajectories.m”)

Supplementary material B: Peak sagittal parameters: comparison with literature

As mentioned before most studies on walking-speed dependencies in joint trajectories mainly focused on the extreme values. The goal of this supplementary material is to provide a quantitative comparison between our findings and previous studies. This is done for all commonly reported extreme values for the knee, hip and ankle joint. Table 7 (see end of this document) summarized the study characteristics of the studies included in this comparison.

Similar to previous studies we observed a speed-dependent increase in maximum hip extension and flexion (Fig. 9 B, D, see end of this document). At higher walking speeds the step size increases (Kirtley et al., 1985; Oberg et al., 1993; Pepin et al., 2003; Van Hedel et al., 2006) (Fig. 12 B), and an increased hip angular range is a requirement to make larger steps. The percentage of the gait cycle at which the maximum hip extension occurs decreased at higher walking-speeds (Fig. 9 A, C), which is related to the reduction of the double support phase at higher walking-speeds (Fig. 12 C).

The maximum knee flexion during the swing phase and the loading response are also well-documented extreme values. The increased knee flexion during the loading response (Fig. 10 B) is suggested to be due to the need for greater shock absorption at higher speeds (Lelas et al., 2003; Pepin et al., 2003). The maximum knee flexion during the swing phase also increased with walking-speed, and occurred earlier in the gait cycle, similar to data reported by several authors (Fig. 11 C, D). The swing phase is often described as “ballistic”, and is achieved passively (Mcgeer, 1990). Consequently, the increased knee flexion during the swing phase is thought to be related to a larger knee angular velocity at toe-off at higher walking-speeds (Anderson et al., 2004).

The ankle joint also contains multiple extreme values that have been studied before. In this study the peak plantar flexion increased with walking-speed and occurred earlier in the gait cycle, which is as also reported in previous studies (Fig. 11 A, B). This increase is suggested to be associated with an increased ankle power, in order to produce adequate push-off at higher speeds (Lelas et al., 2003; Silder et al., 2008; Stoquart et al., 2008). For the peak dorsiflexion during the stance phase we did not find a correlation with walking-speed, whereas others reported a mild reduction (Fig. 11 D). For the peak dorsiflexion during swing we observed a modest reduction with walking-speed, similar to others (Fig. 11 F).

The majority of the studies used in this comparison (Table 7) also provide mean values, or regression equations, for common spatiotemporal parameters like cadence, step length or the relative duration of the different gait phases. Similar to previous studies, the increase in walking-speed is achieved by increases in both cadence and step length (Fig. 12 A, B). Since this was anticipated we calculated the step-ratio. (Sekiya and Nagasaki, 1998) found this parameter to be relatively constant over a wide range of walking speeds. Others also showed a linear relationship between step frequency and step length (Zijlstra et al., 1995). Compared to our results (Sekiya and Nagasaki, 1998) reported higher values for the step-ratio (Fig. 12 A, B). This might be explained by the difference in age. Elderly are known to walk with a higher cadence and shorter step length at similar speeds, resulting in a lower step-ratio (Nagasaki et al., 1996). The regression analysis revealed a small positive dependency of the step-ratio on walking speed and body-height. In contrast, (Zijlstra et al., 1995) reported a reduction in step-ratio at higher speeds. However, their reductions in step-ratio is seen at walking speeds that exceed our maximum speed. A reduction of the step-ratio at extreme speeds may be caused by biomechanical factors limiting a further increase in step length.

At higher walking-speeds the relative double support phase decreased (Fig. 12 C) and, consequently, the swing phase increases. As mentioned above, the swing phase is often assumed to be achieved passively and maintains approximately the same duration at different speeds, leading to an increase in the relative duration at higher walking speeds (where the cycle time decreases). At even higher walking-speeds the double support phase will continue to decrease until the subject starts running and the double support phase becomes absent (Grieve and Gear, 1966).

Supplementary material C: Clinical and robotic applications and limitations

Clinical applications

Although our primary goal was to create reference trajectories that can be used for robotic gait applications they can also serve a clinical purpose. Our results clearly show that walking-speed should be taken into account when judging whether a reduction in joint excursions can be considered pathological. Stroke survivors, for example, often show a reduced hip extension and a reduction in knee flexion during swing (Chen et al., 2005). These patients are also known to walk at reduced speeds. Reversely, when patients, during their rehabilitation process, increase their knee flexion, this could (partly) be explained by their increase in walking-speed. Similarly, in ISCI it has been shown that observed gait alterations are related to both their reduced walking-speed and to their neural deficits (Pepin et al., 2003)

Consequently, gait patterns of neurological patients should be compared with healthy subjects walking at matched speeds in order to differentiate between gait adaptations that are due to a reduced walking-speed and gait adaptations that are a direct consequence of the injury. The reference patterns presented in this study can be used for this purpose. Traditionally, reference patterns are recorded at a limited number of speeds. The obtained regression models in this study allow the reconstruction of reference patterns for any speed, within the given range (0.5-5 kph). This is crucial for the kinematic comparison of patients who are less capable of walking at a gait speed that differs from their 'preferred' or 'comfortable' one (Pepin et al., 2003; Turnbull et al., 1995). During the actual gait therapy, the speed dependent trajectories can also provide the therapist with kinematic references (maximum joint excursions) that are appropriate, and achievable, for the patient (taking into account his current walking-speed).

To employ the provided reference patterns in a meaningful manner a measure of the model error is required. In this study the model error (defined by the RMSE between the model prediction and the actual joint trajectories) is 2.6 degrees (Table 6). This is very close to the inter-subject variability, which is 2.5 degrees (Table 6). This indicates that the model error is equally large as the error that occurs when average trajectories (averaged across subjects) are used. Generally, the largest model errors are found at the lower walking-speeds, and are located during the parts of the gait cycle where the joint excursions are largest, or joint angles change rapidly. At 0.5 kph the RMSE (averaged over the different joint) is 2.8 degrees. Still, this is only marginally larger than the inter-subject variability at that speed (2.7 degrees). At low walking-speeds the subjects also showed more variability in cadence. This indicates that the subjects chose different walking strategies at lower speeds, resulting in variability in their angular trajectories. The agreement between the model error and the inter-subject variability was expected since the reconstructed trajectories greatly resemble the average trajectories (Fig. 8). Thus, for clinical use the reconstructed trajectories can be used in the same way as average trajectories are being used.

Throughout the literature several methods are proposed to quantify gait deviations and detect actual gait deviations (Duhamel et al., 2004; Schutte et al., 2000; Schwartz and Rozumalski, 2008). Here the relatively large inter-subject variability indicates that a patient has to deviate considerable from the average trajectory (or reconstructed trajectories) in order to fall outside the range of natural variability. Still, an actual gait deviation does not necessarily reflect a clinical relevant gait impairment. Consequently, results from quantitative gait data analysis still require the clinician to make an objective assessment of the functional impact of the observed gait alterations. In this process the gait patterns provided in this study can highlight parts of the gait cycle where problems are dominant or determine whether treatment, directed at those problems, show improvements in the right direction.

When the provided reference patterns are used for kinematic comparisons, one should keep in mind that small differences can occur due to different experimenters, gait analysis systems, biomechanical models and testing protocols (Gorton et al., 2009; Schwartz et al., 2004). Still, we believe that these differences are small. Supplementary material B shows kinematic parameters that are reported in different studies, using different protocols, biomechanical models etc. Although there exists some variability, most results fall within the variability observed in our study, indicating that the experimental factors do not affect the generalizability of our results.

Finally, this study shows that one should be careful when comparing mean extreme values and observed extremes in averaged joint trajectories. Most studies present a limited number of specific gait features (often angular extremes) to quantify the effect of different pathologies, whereas other present average joint trajectories. This study shows that these angular extremes usually have a slightly different distribution throughout the gait cycle between subjects, causing lower extreme values in the averaged joint trajectories, compared to the averaged peak values themselves.

Robotic gait applications

Because the model error is in the same order as the natural variability between subjects we concluded that the accuracy of the reconstructed reference trajectories was acceptable. However, for the implementation of these trajectories in robotic gait applications there remain some considerations.

First, some patients might feel hindered when the applied trajectories are slightly different from their own trajectories. Also, applying a “normal gait pattern” might not be feasible (or preferable) for some patients. However, because of the natural variability between subjects it is practically impossible to create a 100% match for every individual patient a priori. Therefore, we consider the reconstructed trajectories as an initial guess of the patient’s walking pattern. When the therapist observes that the reference trajectory is not adequate for the patient, it can easily be modified by changing some key events. For instance; if a patient requires more knee flexion, this can be accomplished by increasing the y parameter of the “max. swing” key event (Fig. 2) and reconstructing a new trajectory. In fact, this “reference pattern tuning” approach is implemented in the control software of a robotic gait trainer that is developed at our department. Currently, this robotic gait trainer (LOPES) is getting installed into two Dutch rehabilitation centers, and therapists have indicated that they greatly value this approach. They also value that the reference trajectories are automatically adjusted to the current walking speed.

Second, robots can potentially harm the user when the reference trajectories (and consequently the robot and user) reach angles that are not within the user’s range of motion. In this study the reconstructed reference trajectories are all within the normal range of motion of the different joints.

Finally, the reference trajectories have to be checked to ensure that there does not exists a combination of angles where the feet will hit the treadmill, or each other. This is more crucial in certain phases (in this case the swing phase). Whether or not the feet will hit the treadmill during the swing phase is a bit more difficult to assess because it depends on the degrees of freedom of the device. If, for instance, the device allows natural pelvis motions (normal lateral pelvic tilt etc.), the patterns are not expected to pose any problems. However, in some robotic gait trainers pelvis motions are restricted or limited. Then, the reference patterns will need a more thorough check before being used. Still, most robotic gait trainers already have (device-) specific safety implementations which will prevent stumbling or other unsafe situations.

Figures

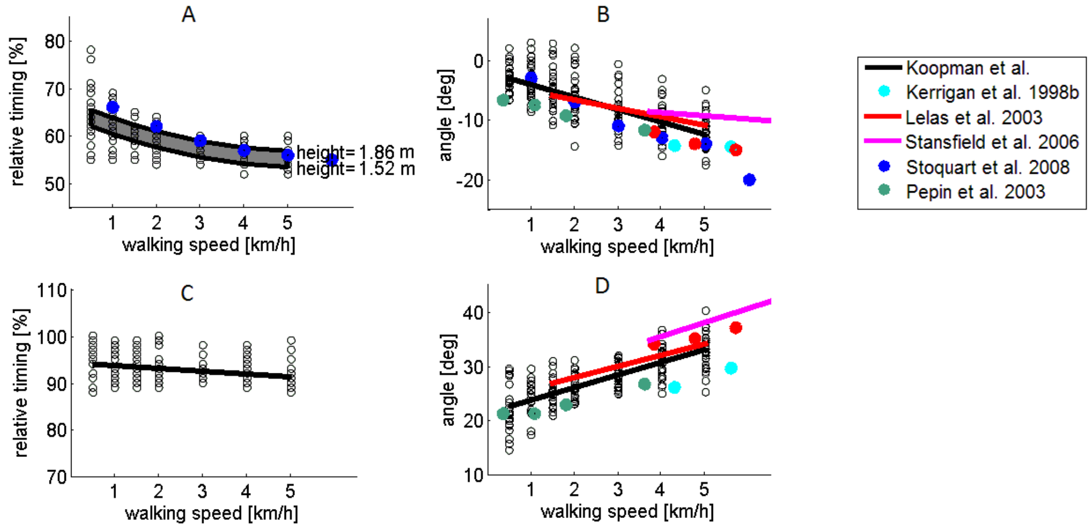

Fig. 9

Comparison with literature for the peak sagittal parameters of the hip joint. Relative timing and amplitude of the maximum hip extension (A, B) and hip flexion (C, D). Each black circle represents the parameter value of the key-event at a specified walking-speed for one subject, whereas the black line indicates the regression model. The shaded area indicates the fitted regression model for the range of body-heights that are included in this study (1.52-1.86 m). The colored lines indicate regression models, and the circles mean values, for different studies.

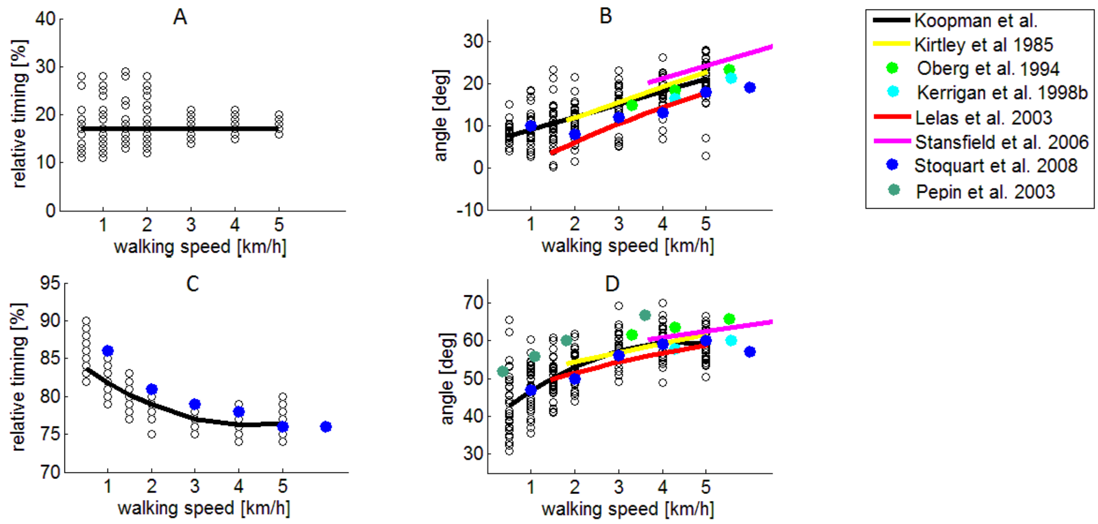

Fig. 10

Comparison with literature for the peak sagittal parameters of the knee joint. Relative timing and amplitude of the maximum knee flexion during the stance phase (A, B) and the swing phase (C, D). Each black circle represents the parameter value of the key-event at a specified walking-speed for one subject, whereas the black line indicates the regression model. The colored lines indicate regression models, and the circles mean values, for different studies.

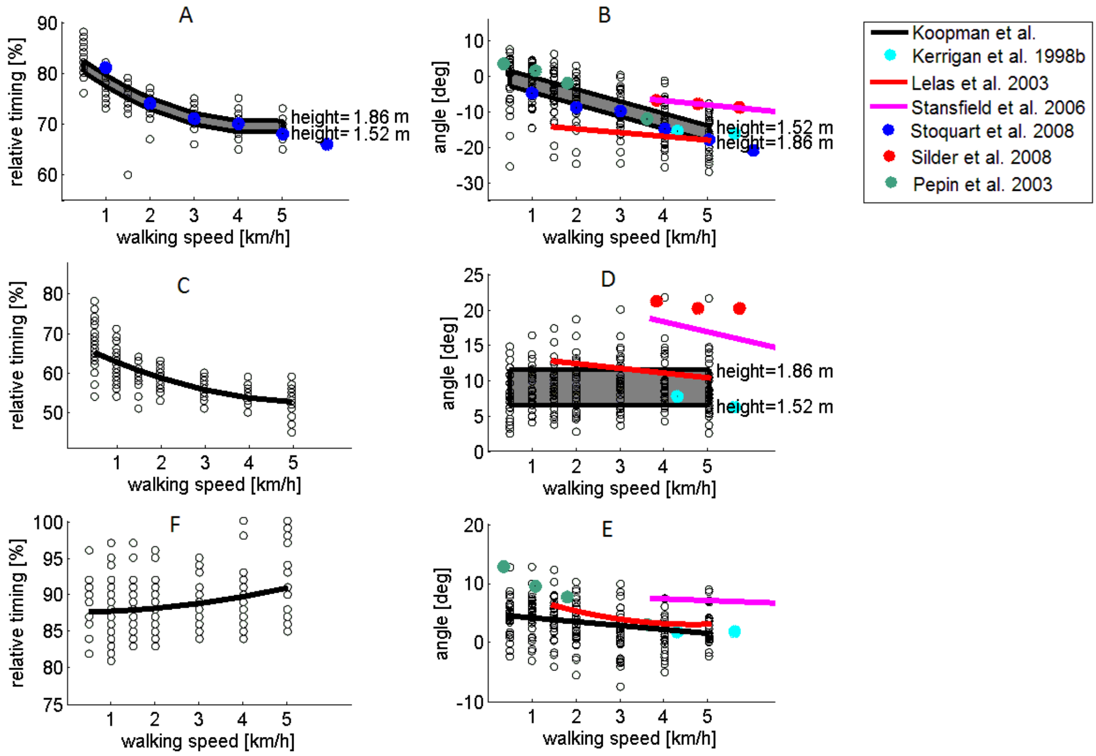

Fig. 11

Comparison with literature for the peak sagittal parameters of the ankle joint. Relative timing and amplitude of the maximum plantar flexion (A, B), maximum dorsiflexion during the stance phase (C, D) and maximum dorsiflexion during swing (E, F). Each black circle represents the parameter value of the key-event at a specified walking-speed for one subject, whereas the black line indicates the regression model. The shaded area indicates the fitted regression model for the range of body-heights that are included in this study (1.52-1.86 m). The colored lines indicate regression models, and the circles mean values, for different studies.

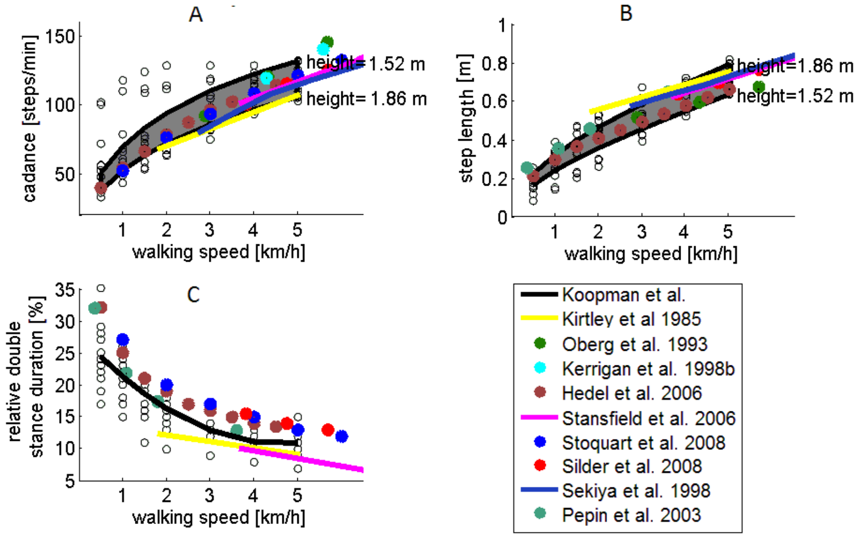

Fig. 12

Comparison with literature for spatiotemporal parameters. Cadence (A), step length (B) and relative double support phase duration (C). Each black circle represents the spatiotemporal parameter at a specified walking-speed for one subject, whereas the black line indicates the regression model. The shaded area indicates the fitted regression model for the range of body-heights that are included in this study (1.52-1.86 m). The colored lines indicate regression models, and the circles mean values, for different studies.

Tables

Table 7

Study characteristics and reported parameters of related studies on speed-dependent gait characteristics.

Study	Type of data	Provided data[1] joints/spatio-temporal data	Age range (mean ± std)	Subjects men-women	Treadmill/ Walkway	Walking speeds[3]	Note
(Kirtley et al., 1985)	Regression lines	Knee flexion Cadence Step length Gait phases	18-63 (37)	10-0	Walkway	Normal [3], 2 below normal, 2 above normal	
(Oberg et al., 1993)	Mean values	Cadence Step length	10-79	116 - 117	Walkway	Slow, normal and fast	Parameters averaged over the age groups 50-59 and 60-69.
(Oberg et al., 1994)	Regression lines Mean values	Hip flexion Knee flexion	10-79	116 - 117	Walkway	Slow, normal and fast	Regression lines are only provided for the total subject population. Values used here are averaged over the age groups 50-59 and 60-69. Corresponding walking-speeds are not reported for the different age groups. The used values are the means of all subjects. For the hip only the ROM is provided, and not the peak flexion/extension separately.
(Kerrigan et al., 1998)	Mean values Mean trajectories	Hip flexion Knee flexion Ankle flexion Cadence	65 - 84 (72.7±5.5)	14 - 17	Walkway	Normal and fast.	Also young subjects included but they only walked at their preferred walking-speed.
(Van Hedel et al., 2006)	Mean trajectories	Cadence Step length Gait phases	19-32 (23.8±3.4)	10 - 10	Treadmill	10 pre-defined speeds (0.5-5 kph)	Mean values for the cadence, step length and gait phases obtained from figure 2. Normalized step length is reported. Here the step length is calculated for a subject with a body-height of 1.69m.
(Lelas et al., 2003)	Regression lines	Hip flexion Knee flexion Ankle flexion	19 -40 28.1±5.9	64 (un-known gender)	Walkway	Normal, fast, slow and very slow	

(Stansfield et al., 2006) [2]	Regression lines	Hip flexion Knee flexion Ankle flexion Cadence Step length Gait phases	7 – 12	8 - 8	Walkway	Normal	Prospective 5-year study. Regression formulas are expressed in normalized speed. The reported spatiotemporal parameters are also normalized to dimensionless quantities. Here the parameters are calculated for a subject with a body-height of 1.69m.
(Stoquart et al., 2008)	Mean values Mean trajectories	Hip flexion Knee flexion Ankle flexion Cadence Gait phases Timing of key features	23±2	4 - 8	Treadmill	6 pre-defined speeds (1-6kph)	
(Silder et al., 2008)	Mean values	Hip flexion Ankle flexion Cadence Step length Gait phases	Young 18–35 Elderly 65 - 85	Young 9 - 11 Elderly 7 - 13	Walkway	80%, 100%, and 120% of normal	Only the data from the elderly is used for comparison. Normalized step length is reported. Here the step length is calculated for a subject with a body-height of 1.69m.
(Pepin et al., 2003)	Mean values Mean trajectories	Hip flexion Knee flexion Ankle flexion Cycle time Step length Gait phases	28 -40	7-0	Treamill	0.1, 0.3, 0.5 and 1.0 m/s	
(Sekiya and Nagasaki, 1998)	Mean values	Step-ratio	Males 22.4±3.7 Females 22.5±3.9	8-17	Walkway	Slowest, slow preferred fast, fastest	Step-ratio used to calculate step length and cadence. Values are averaged over male and females.
Koopman	Regression lines	Hip abduction Hip flexion Knee flexion Ankle flexion Cadence Gait phases	47-69 (59.4±6.3)	7 - 8	Treadmill	7 pre-defined speeds (0.5-5 kph)	

[1] Only the reported data that is used for the comparison with this study is mentioned in the table.

[2] Although this study present data for growing children it is shown that the change in self-selected speed, not age, is the primary determinant of kinematic and kinetic changes observed in growing children and elderly (Kerrigan et al., 1998; Riley et al., 2001; Stansfield et al., 2001; van der Linden et al., 2002). Therefore the provided normalized parameters (and walking-speed) are scaled to a subject with a body-height of 1.69m.

[3] Normal refers to the self–selected walking-speed.

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